



DOCTOR OF MEDICINE

A biomechanical evaluation of a novel intramedullary device for hip fracture fixation

Page, Piers

Award date:
2019

[Link to publication](#)

Alternative formats

If you require this document in an alternative format, please contact:
openaccess@bath.ac.uk

Copyright of this thesis rests with the author. Access is subject to the above licence, if given. If no licence is specified above, original content in this thesis is licensed under the terms of the Creative Commons Attribution-NonCommercial 4.0 International (CC BY-NC-ND 4.0) Licence (<https://creativecommons.org/licenses/by-nc-nd/4.0/>). Any third-party copyright material present remains the property of its respective owner(s) and is licensed under its existing terms.

Take down policy

If you consider content within Bath's Research Portal to be in breach of UK law, please contact: openaccess@bath.ac.uk with the details. Your claim will be investigated and, where appropriate, the item will be removed from public view as soon as possible.

A biomechanical evaluation of a novel intramedullary device for hip fracture fixation

Volume 1 of 1

Piers Robert John Page

A thesis submitted for the degree of Doctor of Medicine

University of Bath

Department for Health

December 2018

COPYRIGHT

Attention is drawn to the fact that copyright of this thesis rests with the author. A copy of this thesis has been supplied on condition that anyone who consults it is understood to recognise that its copyright rests with the author and that they must not copy it or use material from it except as permitted by law or with the consent of the author.

This thesis may be made available for consultation within the University Library and may be photocopied or lent to other libraries for the purposes of consultation with effect from

Signed on behalf of the Department for Health

Table of contents

A biomechanical evaluation of a novel intramedullary device for hip fracture fixation	2
Table of contents	3
List of figures.....	6
List of tables.....	8
Abbreviations	9
Contributions by others	11
Acknowledgements.....	12
Abstract	13
Introduction	13
Methods.....	13
Results	14
Discussion	14
Introduction.....	16
Chapter 1: Hip fractures – the clinical landscape.....	18
1.1 Hip fractures.....	19
1.11 The nature and scale of the problem	19
1.12 Causes of hip fracture	20
1.13 Classification.....	22
1.2 Bone	25
1.21 Structure	25
1.22 Healing	27
1.23 Osteoporosis.....	28
1.3 Management of hip fractures	30
1.31 The rationale for operative hip fracture care.....	30
1.32 History of surgical interventions	31
1.33 Current standard of care devices	36
1.4 Current clinical evidence for fixation strategies	39
1.41 Evidence from existing meta-analysis	39
1.42 Recent updates to the evidence	41
Summary of key clinical points in the literature	47
Chapter 2: Biomechanics	48
2.1 Biomechanics of the native hip	49
2.2 Biomechanics of hip fractures and their fixation	52
2.3 Design considerations for fixation devices	55
2.31 Intramedullary nails	55

2.32	Rotational stability	57
2.33	Torsional stability	61
2.34	The X-Bolt system	63
2.35	Metallurgy	65
2.41	Bone and substitutes.....	66
2.42	Simulating physiological forces at the hip.....	67
2.43	Computer-based modelling	68
2.5	Summary of key biomechanical points in the literature.....	70
Chapter 3: An evaluation of incidence and aetiology of failure of fixation in extracapsular hip fractures		72
3.1	Introduction	73
3.2	Patients and Methods	75
3.21	Patients	75
3.22	Sampling	76
3.23	Methods.....	77
3.3	Results	79
3.31	Patients by site	79
3.32	Patient characteristics.....	80
3.33	Surgical characteristics	82
3.4	Discussion	86
Chapter 4: A comparison of 3-point bending in stainless steel and titanium alloy nails		89
4.1	Aims and objectives	90
4.2	Materials and models	92
4.21	Materials.....	92
4.22	Construction of the synthetic bone jig	94
4.23	Creation of the uniform fracture model	95
4.24	Creation of the instrumented femur model	97
4.3	Methods.....	99
4.31	Testing.....	99
4.32	Calculations and statistical methods	101
4.4	Results	103
4.5	Discussion	107
Chapter 5: A comparison of resistance to torsion between the X-Bolt device and a conventional compression screw.....		110
5.1	Aims and objectives	111
5.2	Materials and models	112
5.3	Methods.....	113
5.4	Results	116

5.5	Discussion	119
Chapter 6:	Discussion	122
6.1	Key findings and their context	123
6.11	Clinical study	123
6.12	Biomechanical studies.....	125
6.13	The paucity of evidence behind the introduction of orthopaedic implants for trauma	126
6.2	Strengths and limitations.....	127
6.21	Clinical considerations	127
6.22	Femoral model	130
6.23	Experimental methodology	132
6.3	Methodological lessons learned	135
6.4	Conclusions	137
6.5	Future research directions	139
Bibliography.....		141
Appendices.....		158
Appendix 1 – Literature review search strategy		159
PubMed:		159
EMBASE		160
CENTRAL		161
Appendix 2 – Full torsional testing data by model		163

List of figures

Figure 1.1 – The AO/OTA classification system for extracapsular hip fractures	24
Figure 1.2 – Fracture of the cervix femoris	31
Figure 1.3 – The Smith-Petersen nail	32
Figure 1.4 – The Hawley plate	33
Figure 1.5 – The Jewett nail plate.....	34
Figure 1.6 – the sliding hip screw (a) in situ and (b) being augmented with a trochanter stabilizing plate.....	36
Figure 1.7 – A modern short intramedullary nail.....	37
Figure 2.1 – Forces acting through the hip in stance	50
Figure 2.2 – (a) Calculation of the tip-apex distance (b) Total screws placed and number cutting out, by zone of femoral head	54
Figure 2.3 – Augmentation of a SHS with derotation screw	58
Figure 2.4 – The InterTAN intramedullary nail with dual-screw system.....	59
Figure 2.5 – The biomechanics of torsional stability	61
Figure 2.6 – The X-Bolt hip fracture fixation device	63
Figure 2.7 – A FEA model of a hemiarthroplasty implant	68
Figure 3.1 – Participant flow diagram.....	79
Figure 3.2 – Type of fixation in (a) all cases and controls (b) all cases and controls with A3 fractures and (c) cases with A3 fractures	84
Figure 4.1 – Gamma nail with (a) failed distal locking and (b) subcapital fracture after removal of compression screw	90
Figure 4.2 – Schematic drawings of (a) Affixus and (b) X-Bolt nails	93
Figure 4.3 – The 31A2 osteotomy jig	94
Figure 4.4 – Distal femoral resection in (a) Postero-anterior and (b) lateral plane	95
Figure 4.5 – Sawbone in jig post-osteotomy	95
Figure 4.6 – Sawbone post-osteotomy, with intermediate lesser trochanter fragment discarded.....	96
Figure 4.7 – X-Bolt nail locked (a) proximally and (b and c) distally.....	97
Figure 4.8 – Orthobone model showing common mode of failure, a periprosthetic fracture in the proximal third of the femur	98
Figure 4.9 – Femoral model mounted for testing of 3-point bending	99
Figure 4.10 – Loading cycle for stiffness testing of femoral models (Extension in mm, load in N)	101
Figure 4.12 – Stiffness of model by group, with single un-instrumented femur for comparison.....	106
Figure 5.1 – Electromechanical testing setup showing (a) acetabular component mounted in jaws (b) specimen engaged in tester and (c) schematic of testing apparatus.	113
Figure 5.2 – Loading cycle for torsional testing of femur models	114

List of tables

Table 1.1 – Recent RCTs relating to the fixation of intertrochanteric fractures ...	42
Activity Table 2.1 – Forces and moments acting through a total hip replacement implant in a live patient during activities of daily living.....	51
Table 3.1 – Patient characteristics by cohort.....	81
Table 3.2 – Surgical characteristics by cohort	83
Table 4.1 – Change in extension between 250N and 450N of load, with calculated stiffness values.....	104
Figure 4.11 – Stiffness of model by group, without un-instrumented femur.....	105
Table 4.2 – Shapiro-Wilk tests for normality of distribution of stiffness in X-Bolt and Affixus models.....	105
Table 4.3 – Independent Samples Mann-Whitney U Test for stiffness of X-Bolt and Affixus models.....	105
Table 4.4 – Independent Samples Kruskal-Wallis Test for stiffness of X-Bolt and Affixus models, with single un-instrumented femur for comparison.....	106
Table 5.1 – Mean displacement in stance and sitting tests by specimen.....	117
Table 5.2 – Shapiro-Wilk tests for normality of distribution of stance and sitting displacement in X-Bolt and Affixus models.....	117
Table 5.3 – Independent Samples Mann-Whitney U testing of standing and sitting displacement between X-Bolt and Affixus models	118
Table 5.4 – Independent Samples Kruskal-Wallis testing of stance and sitting displacement between X-Bolt and Affixus models and un-instrumented femur	119

Abbreviations

AO	Arbeitsgemeinschaft für Osteosynthesefragen (Association for the study of internal fixation)
AP	Antero-posterior
BMD	Bone mineral density
CI	Confidence interval
CT	Computed tomography
DHS	Dynamic hip screw
EQ-5D	Euro-QoL 5 Dimension
FEA	Finite element analysis
HR-pQCT	High resolution peripheral quantitative computed tomography
IMHS	Intramedullary hip screw
IMN	Intramedullary nail
NHFD	National Hip Fracture Database
NHS	National Health Service
NICE	National Institute for Health and Care Excellence
OTA	Orthopaedic Trauma Association
PACS	Picture Archiving and Communication System
PFNA	Proximal Femoral Nail Anti-rotation
RR	Relative risk
RSA	Radiostereometric analysis
SD	Standard deviation

SHS	Sliding hip screw
UK	United Kingdom
USA	United States of America
vBMD	Volumetric bone mineral density

Contributions by others

The additive manufacturing process for the jig was undertaken by Dr. Alisdair McLeod, University of Bath.

Local data in support of the clinical project was gathered by Mr. Niraj Vetharajan, Mr. Adam Smith, Mr. Luke Duggleby and Mr. Michael Field, fellow orthopaedic trainees.

Acknowledgements

Devices for testing were generously provided by Zimmer Biomet and X-Bolt Orthopaedics. X-Bolt Orthopaedics also provided the synthetic femora for all testing.

I would like to thank Dr. Alisdair McLeod for his advice and support during biomechanical testing, and all the workshop team at the Centre for Orthopaedic Biomechanics at the University of Bath for their hands-on assistance in manufacturing and testing.

I am indebted to Adam Smith, Luke Duggleby, Niraj Vetharajan and Michael Field for their tireless efforts in data collection.

My supervisors, Dr. Dario Cazzola, Mr. Michael Whitehouse and Prof. Richie Gill have provided wisdom, support, contacts and a listening ear throughout this project. My gratitude is profound, heart-felt and beyond expression here.

Finally my wife, Caroline and son, William – this whole adventure has been enabled and supported by you, and is dedicated to you.

Abstract

Introduction

Hip fractures are common and disabling injuries, affecting mainly older adults. Around 65,000 patients per year are affected in the United Kingdom (UK), with an estimated economic cost in excess of £1bn . Surgical strategies can be based on either fixation or replacement and, when fixing the joint, two broad classes of device predominate. These are the sliding hip screw (SHS) and the intramedullary nail (IMN). The intramedullary nail is typically indicated in less stable fractures, where its design features help maintain the position of the proximal femur as it heals.

Recently, a new device has been brought to market which can work in either of these configurations, paired with a plate or an intramedullary nail, aiming to give better purchase in the femoral head. The intramedullary device accompanying it is made from stainless steel. Data is available in the National Hip Fracture Database (NHFD) to quantify early failures of fixation, amongst other major complications, but later complications are poorly quantified.

This study sought to quantify those problems which may already appear in the NHFD and those later or unmeasured ones which may not, to compare the stiffness of femoral models with hip fractures fixed with the new stainless steel intramedullary device with those employing a titanium standard of care device, and to compare the resistance to torsional displacement of the femoral head fragment between the novel and standard of care devices.

Methods

Three years' worth of NHFD data were obtained from three sites. All intertrochanteric fractures were identified, and from this cohort any patients with further episodes of care related to this hip fracture were identified, and the reason recorded. These patients were then age- and sex-matched 3 to 1 with problem-free controls. Their fracture classification, tip-apex distance, co-morbidities and cognitive function were recorded, as were their pre-injury mobility levels. Comparisons were then made between problem and problem-free cohorts. Synthetic femora were used for testing of the fixation devices, with instrumented

bones subject to 500 N loading in an electromechanical testing device and 3-point bending recorded. A novel testing device was then created to induce anterior or posterior force in the head of a synthetic femur in which an unstable intertrochanteric fracture had been created, before being subsequently fixed using either the novel device or a standard of care compression screw. The displacement about the longitudinal axis of the femurs was measured in order to quantify the torsional stability of the fixation with each construct.

Results

A total of 4010 patients were entered in the NHFD across three sites between January 2013 and December 2015. Of these, 1260 had sustained intertrochanteric fractures, 57 of whom subsequently went on to experience problems and be referred to hip surgeons. The most common problem was failure of fixation, occurring in 22 patients. There was no difference in age or sex between cohorts, but the problem cohort had higher cognitive capability ($p=0.0300$) and lower co-morbidity as reflected in their American Society of Anesthesiologists (ASA) score ($p < 0.0001$). Distribution of fracture type was the same between both groups, but a higher proportion of highly unstable fractures were fixed with a sliding hip screw in the problem group. The tip-apex distance showed a significant difference between groups ($p=.0050$).

When considering 3-point bending, the stiffness of both constructs was similar at 717.92 N/mm for the model fixed with the stainless steel X-Bolt nail and 729.53 N/mm for that fixed with the titanium Affixus nail ($p=1.0000$).

Resistance to torsional displacement was increased in the novel device ($p < 0.0001$).

Discussion

This thesis sets the context for early clinical trials of the novel X-Bolt nail, a device for the fixation of unstable and more failure-prone extracapsular hip fractures, manufactured from stainless steel and employing a novel expanding bolt device for fixation in the femoral head.

A population of patients exists which has problems after fixation of hip fractures, and these appear to be increased in the group with extramedullary fixation of unstable fractures. The incidence of these problems in this study was 4.5% of patients with some form of problem, with 1.7% of fixed fractures failing. This is in keeping with current rates seen in larger datasets. Put in the context of more than 20,000 such fractures fixed annually, there is an argument which supports the ongoing use of intramedullary devices in this situation. The fact that some failures were seen in the group of patients who underwent fixation with IMNs illustrates that the current generation of devices still exhibit some failures of fixation, and so there is potential room for improvement.

In biomechanical terms, these experiments have demonstrated firstly that stainless steel intramedullary nails can be manufactured to have similar stiffness to titanium ones, and hence this metal should not be discounted for implant manufacturing on this basis alone. The choice of this metal for the X-Bolt nail should not, therefore, preclude clinical testing.

Further, this work has shown that there is a potential benefit in the novel compression bolt, in that in early biomechanical tests it has shown superior resistance to torsional displacement simulating loading of the flexed hip. This loading is a highly important physiological force acting through the hip in many activities of daily life, and hence a device reducing the risk of resultant failure of fixation merits further investigation. Further, clinical work should be considered.

Introduction

Hip fractures are common and debilitating injuries, occurring predominantly in older people, and they represent one of the most common surgically treated injuries in orthopaedic practice. The surgical treatment may be by either replacement of all or part of the joint, or by fixation of the broken bone to stabilize it while it heals. A number of different devices are available to achieve this fixation – in contemporary practice, they tend to be variants of a screw which passes through the femur, along the femoral neck and into the femoral head, with either a supporting plate attached to the side of the femur, or a metal rod or nail which is implanted into the canal of the femur itself, with the screw passing through it.

This thesis describes a process of pre-clinical validation of a novel device for the fixation of hip fractures, the X-Bolt nail. This employs an expanding bolt which passes through a nail implanted within the canal of the femur, which the manufacturer claims improves the stability of the surgical construct.

The thesis explains the injuries and their importance for those who sustain them and goes on to describe key underpinning biomechanical principles of both the injuries themselves and of the various strategies and devices used to treat them.

The first experimental part of the thesis reports a retrospective case-control cohort study which describes the populations undergoing fixation of their hip fractures in three different hospitals in the south of the United Kingdom, looks at what implants were used to fix them and compares rates of failure between different sub-types of fracture and types of implant. It aims to demonstrate that some subtypes of fracture may be better fixed by IMN than by SHS, by showing differential rates of failure of fixation and, furthermore, that failures even in the IMN group demonstrate a potential unmet need for a further evolution of the class of device.

A biomechanical sub-study then compares stiffness of the experimental device with that of one in common use already in the National Health Service, as the experimental nail is made from stainless steel, by contrast with the titanium used in most similar devices. Many surgeons have, anecdotally, expressed reluctance to

use a stainless steel device as early failures were seen when previous generations of this type of device were made from steel. This experiment provides proof of concept that modern orthopaedic engineering principles can overcome these early flaws.

A further biomechanical sub-study examines the resistance to torsion (rotation of the femoral head fragment about the long axis of the femur) of the novel expanding bolt, which replaces the screw conventionally used in many such devices. Failure of fixation often occurs through the device cutting out; not maintaining its position as implanted. A device which is more resistant to this torsional displacement is, therefore, very attractive to hip fracture surgeons. The experiment described in this thesis tests this property by simulating loads at the hip experienced in everyday life, such as in sitting down, standing up or climbing stairs.

The overall purpose of this doctoral project is to demonstrate the need for intramedullary nails, the class of device to which the X-Bolt nail belongs, to show that stainless steel nails can safely be implanted and that there is a potential benefit to the novel expanding bolt. This pre-clinical validation should then create the context for clinical testing of the device, a scientifically and ethically essential step before considering implantation of this device in living patients.

Chapter 1: Hip fractures – the clinical landscape

1.1 Hip fractures

1.11 *The nature and scale of the problem*

More than 65,000 patients sustained a hip fracture in England, Wales or Northern Ireland last year (Royal College of Physicians, 2017). Their hospital treatment and rehabilitation alone has been estimated to cost the British economy in excess of £1 bn annually (Leal et al., 2015). It has been suggested that a global trend of increasing incidence is likely for several years to come; people are living longer and, with a median age of 85 years at time of injury, patients with hip fractures are going to feature more prominently in any society where longevity increases (Hernlund et al., 2013).

The effect of these injuries is profound – historically, a one-year mortality of around one-third of patients was described, with one-third of these deaths occurring within the first month post-injury (Keene et al., 1993; Jensen & Tøndevold, 1979). The traditional approach to understanding these injuries has been one focused on defined surgical, health economic and healthcare outcomes such as mortality, cost and length of stay. A more modern, patient-centred understanding is now emerging and health related quality of life as measured by instruments such as the EuroQuOL Five Dimension (EQ5D) questionnaire is becoming widely accepted as a key outcome (Parsons et al., 2014). Such research paints a stark picture – in a rigorous and contemporary study, patients have been shown never to recover their previous function after hip fracture and, for a number of months post-injury, rate their health state as worse than being dead (Griffin et al., 2015).

Aside from the deleterious effects on the individual, there is a self-evident impact on the health economy if a society's more vulnerable members become less independent and are no longer able to live in their own homes. Only 10% of patients featured in the 2017 National Hip Fracture Database report described themselves as able to walk without an aid at 4 months post-injury, which may give a clearer idea of the support potentially required in activities of daily living than the 67% of patients reported to have returned to their original residence by this point (Royal College of Physicians, 2017).

1.12 *Causes of hip fracture*

Hip fractures are commonly described as “fragility fractures” or “osteoporotic fractures” but the pathophysiology is, in fact, more complex. The reduction in bone mineral density pathognomonic of osteoporosis contributes to a fracture-prone state, but a propensity to fall also contributes substantially to hip fracture. The latter tends to be multi-factorial – states of impaired balance from peripheral or central neurological problems, reduced mobility resulting from arthritic joints and a lessened ability for self-care, leading to a home environment where obstacles or hazards may go unnoticed can all be part of this. Wagner’s prospective, longitudinal study of 24,598 Swedish twins found that in the 2,890 who were non-concordant in their answer to the question “Do you have impaired balance?”, there was an odds ratio of 3.13 (95% confidence interval (CI) 1.62 to 6.05) of hip fracture against the twin answering affirmatively to balance problems, predicting 40% of hip fractures (Wagner et al., 2008). This finding adds context to Stone et al.’s longitudinal cohort study of 9,704 women over the age of 65 years in the United States of America (USA), which investigated the relationship between bone mineral density at several sites and fracture risk, reporting that a bone mineral density measurement (as measured by dual energy X-ray absorption or DEXA) ≤ 2.5 standard deviations below the norm in the femoral neck was associated with a proportion of 0.28 (95% CI 0.22 to 0.33) of hip fractures attributable to osteoporosis (Stone et al., 2003). A patient’s sense of balance could therefore be argued to be as important a factor as their bone density when considering primary prevention of hip fractures.

Independently of the altered bone mineral density seen in osteoporosis, altered bone microarchitecture may also be implicated. Highly focal alterations in cortical thickness and trabecular microstructure can contribute to risk of fracture at a given site, and a multi-centre high resolution peripheral quantitative computed tomography (HR-pQCT) study of 1,379 women found reduced cortical thickness and a reduced number of trabeculae in those with fractures, findings which were independent of their total hip T-score (Boutroy et al., 2016). This study also found altered volumetric bone mineral density (vBMD) deficits with the same association, implying that vBMD measurement may offer a more predictive value than its DEXA counterpart.

Dargent-Molina et al.'s multicentre study of 7,575 women over the age of 75 in France found four factors to be independently predictive of hip fracture, in addition to BMD, when analyzed in a Cox regression model (Dargent-Molina et al., 1996). These factors were low gait speed (relative risk (RR) increase of 1.4, 95% CI 1.1 to 1.6 for ever 1 SD reduction), difficulty in tandem heel-to-toe walk (RR 1.2, 95% CI 1.0 to 1.5 for each point on difficulty scale), visual acuity (RR 2.0, 95% CI 1.1 to 3.7 for acuity $\leq 2/10$) and calf circumference (RR 1.5, 95% CI 1.0 to 2.2). This demonstrates that sight, neuromuscular function and muscle mass all contribute to an impaired locomotor function and hence risk of hip fracture.

On the basis of such evidence, authors such as Järvinen et al. (Järvinen et al., 2008) argue that hip fracture prevention strategies should avoid undue emphasis on pharmacological interventions to address BMD and prioritize instead approaches which also promote physical exercise regimes and the use of fall protectors.

While an understanding of the multifactorial causes of hip fracture is important, it is also important that the surgical strategies employed acknowledge the fact that there is a very high prevalence of altered bone quality in this population that must be addressed with any surgical intervention to treat the hip fracture, a central consideration in the experiments contributing to this project.

1.13 *Classification*

The classification of hip fractures is, broadly, an anatomical one based on the blood supply to the femoral head. This supply is primarily via several vessels running along the femoral neck, within the capsule of the hip joint and via intramedullary vessels also disrupted by the fracture, with the intra-articular ligamentum teres containing only some smaller vessels and hence contributing only a portion of the required vascular supply. Fractures are grouped into those which are intracapsular, inside the capsule, and the extracapsular ones occurring outside it. Intracapsular fractures are presumed, unless undisplaced, to have caused vascular injury and hence to be at high risk of having devascularized the femoral head. In the elderly patient group in whom the majority of these injuries occur, this vascular injury has historically been presumed to render the native femoral head unsalvageable and hence to mandate some form of prosthetic replacement. The actual incidence of failure of fixation has been reported to range between 10% and 60%, with a proportion of these being those who go on to non-union rather than avascular necrosis (Loizou & Parker, 2009; Davison et al., 2001; Gjertsen et al., 2011). Extracapsular fractures, by contrast, usually retain the blood supply to the femoral head and so lend themselves to fixation.

A number of classifications exist. The Garden classification is a radiological system describing the extent and displacement of the fracture: type I fractures are incomplete or valgus impacted fractures; type II complete but undisplaced; type III complete and partially displaced; type IV complete and completely displaced (Garden, 1964). Pauwel's classification assessed how vertical the fracture line was, with an increased verticality presuming higher shear force across the fracture and hence higher likelihood of failure of the fixation (Pauwels, 1935). More recent work has demonstrated that neither classification accurately predicted union of these fractures (Parker & Dynan, 1998). It is also salient that, as treatment has moved away from fixation in the majority of these injuries, these classifications themselves predict failure of a treatment now not routinely used for intracapsular fractures because of the risk of failure (National Institute for Health and Care Excellence, 2017).

In modern practice, the AO/OTA classification (Figure 1.1) offers more prognostic utility and treatment guidance (Müller et al., 2012). This system

classifies all fractures of long bones by anatomical location, proximal or distal, type and subtype of fracture into an alphanumeric code. Hip fractures are coded as 31A or 31B fractures for extracapsular and intracapsular subtypes respectively. The 31A, extracapsular type is then divided into simple 2-part pertrochanteric 31A1 fractures, 31A2 multifragmentary pertrochanteric fractures and 31A3 intertrochanteric fractures, the stability of the fracture pattern decreasing as the final digit increases. While no large studies exist, multiple smaller ones have shown good intra- and inter-observer reliability in this classification system (Pervez et al., 2002; Schipper et al., 2009).

Subgroups and Qualifications:

Femur, proximal, pertrochanteric simple (only 2 fragments) (31-A1)

1. Along intertrochanteric line (31-A1.1)

2. Through the greater trochanter (31-A1.2)

(1) nonimpacted
(2) impacted

3. Below lesser trochanter (31-A1.3)

(1) high variety, medial fracture line at lower limit of lesser trochanter
(2) low variety, medial fracture line in diaphysis below lesser trochanter

A1



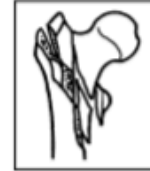
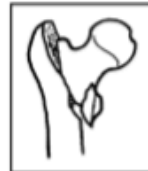
Femur proximal, trochanteric fracture, pertrochanteric multifragmentary (always have posteromedial fragment with lesser trochanter and adjacent medial cortex) (31-A2)

1. With 1 intermediate fragment (31-A2.1)

2. With several intermediate fragments (31-A2.2)

3. Extending more than 1 cm below lesser trochanter (31-A2.3)

A2



Femur, proximal, trochanteric area, intertrochanteric fracture (31-A3)

1. Simple oblique (31-A3.1)

2. Simple transverse (31-A3.2)

3. Multifragmentary (31-A3.3)

(1) extending to greater trochanter
(2) extending to neck

A3

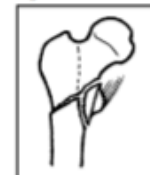
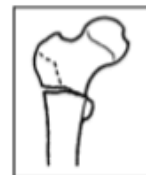


Figure 1.1 – The AO/OTA classification system for extracapsular hip fractures
(Orthopaedic Trauma Association Classification, Database and Outcomes Committee, 2007)

Since the inception of this project, the AO / OTA classification system has been revised, with the 31A group undergoing some changes (Kellam et al., 2018; OTA Classification, Outcomes and Database Committee, 2018). This work is based on the system in use at the time of inception of the project.

1.2 Bone

1.21 *Structure*

Bone is a composite tissue comprising a matrix of hydroxyapatite and collagen, a connective tissue protein, and a number of cells. It takes two primary forms, woven (immature) and lamellar (mature). These forms differ by way of organization of constituent parts, cell and water content. Woven bone is seen in much higher proportions in children and, with the exception of a limited number of sites and structures, is not seen beyond the age of four years except in the process of fracture healing (Buckwalter et al., 2010).

Woven bone features randomly orientated collagen fibrils and an irregular pattern of mineralization. It has a relatively higher cell count and water content. This makes it weaker than lamellar bone, but more flexible, and permits it to behave isotropically in response to deformation (its mechanical properties remain constant, regardless of the orientation of the force applied). Lamellar bone, by contrast, features tightly-packed, organized layers, where fibrils run parallel in a direction. Layers are oriented perpendicularly to each other as they stack up, and feature connections within and between layers, resulting in a pattern similar to the orientation of the grain in plywood (Rho et al., 1998).

Within the matrix, cells of three primary types are found. Osteoblasts are mononuclear cells responsible for the formation of new bone, via the production of osteoid which subsequently mineralizes. When active, they contain large numbers of endoplasmic reticulae, mitochondria and Golgi apparatus and take a rounded profile. An osteoid “seam” forms between the cell border and the mineralized matrix, perforated by cytoplasmic projections which reach the osteocytes in the mineralized bone, forming what is thought to be a control mechanism. At the end of their active phase, their fate is to flatten and become a bone-lining cell, to surround themselves with matrix and become an osteocyte, or to disappear.

Osteocytes are surrounded by organic matrix, which can subsequently mineralize, and form the majority of the cellular population in bone. They have cytoplasmic projections, reaching through canaliculi to other cells, and so have means of co-ordination. In particular, this is thought to play a part in the exchange of minerals

between extracellular fluid and bone. They have a single nucleus and a volume and distribution of organelles which varies in line with their activity at any given time.

Osteoclasts have a resorptive function and a key role in remodelling bone. They derive from the monocyte and macrophage cell line and, as inactive preosteoclasts, can be found in marrow and peripheral blood. In their active form after stimulation, they multiply and form larger, multinucleate osteoclasts with 3-20 nuclei. They are rich in mitochondria and lysosomes, providing the energy and substrate to drive the resorptive function they fulfil. On a microscopic level, they can be seen to form a ruffled or “brush” border, maximizing the surface area with which they resorb the mineralized matrix. Rarely seen in normal bone, they may be characterized by Howship’s lacunae, indentations in the surface to which they are attached.

1.22 *Healing*

Bone healing is a continuum of physiology which may take different forms at different stages in the process, dependent on a number of patient and injury factors. At the time of injury, a process of inflammation begins; a haematoma forms at the fracture site and with it arrives a number of precursor cells.

Meanwhile, mesenchymal cells migrate to the site and differentiate into osteoblast or chondrocyte lines, providing the raw materials for the endochondral phase of healing. The fracture gap is bridged by chondrocytes which form a soft callus of fibrocartilage. This soft callus mineralizes to hard callus, at this stage structurally similar to woven bone. While this happens, intramembranous formation of bone occurs on the periosteum itself – this adds little in terms of strength, but provides an additional scaffold around which healing can continue. Further remodelling then occurs, by the balanced action of osteoblasts and osteoclasts, until normal architecture returns.

This process of secondary bone healing is one which happens in nature without intervention. In primary bone healing, a near-perfect reduction is achieved, followed by compression through rigid internal fixation. This obviates the fracture gap, and so healing can occur through the action of osteoclastic cutting cones, termed primary (osteonal) healing.

The manner in which healing occurs can be explained by Perren's strain theory. (Perren, 1979) Strain is a unit-less measurement of change in length of a material when a force is applied to it. Different materials have different tolerances to strain, and this is true of the spectrum of bony substances by which a fracture may heal. Woven bone, for example, is less tolerant of strain than fibrocartilage, and so if a fracture is fixed in such a way that a high-strain environment is created, woven bone may be unable to bridge the fracture gap without failing under strain, whereas fibrocartilage will remain intact. As the fracture healing progresses, the stiffness of the construct increases and the strain reduces, hence less strain-resistant, harder callus can develop (Perren, 2002; 1979).

1.23 *Osteoporosis*

Osteoporosis is “a skeletal disorder characterized by compromised bone strength predisposing a person to an increased risk of fracture” (Lorentzon & Cummings, 2015). It is characterized by a density in L2-L4 vertebrae, as measured on dual energy X-ray absorptiometry (DEXA), or T score, of 2.5 or more standard deviations less than that of a healthy premenopausal woman. There exists also a “pre-osteoporotic” state of osteopenia, defined by a T-score between 1.0 and 2.5. It should be noted that these thresholds for diagnosis are simply points on a distribution curve, and there exists a proportional relationship between BMD and fracture risk (Kanis, 2002; Kanis et al., 1994).

Although historically divided into types I (post-menopausal, leading to increased risk of vertebral fractures) and II (age-related, associated with hip fracture) disease, after the work of Riggs and Melton, this distinction has latterly been suggested to be misleading. Several studies have demonstrated prediction of both hip and spine fractures by low circulating oestradiol, a post-menopausal state, and a peak risk of vertebral fracture at 75 years of age, rather than an increased but unchanging risk post-menopause (Cummings et al., 1998; Roy et al., 2003).

The fractures most commonly associated in the literature with osteoporosis are those of the hip, the distal radius and of vertebrae. It has, however, been demonstrated that the incidence of most types of fracture is increased in those with reduced bone density, and that fracture is one of the strongest predictors of subsequent fracture (Cummings & Melton, 2002).

The management of osteoporosis has focused on identifying those with or at risk of this reduced bone density and offering multimodal advice and therapy to increase it. Such measures include advice on achieving an adequate dietary intake of calcium and vitamin D, exercise regimes, smoking cessation and alcohol consumption guidance and falls prevention measures, including home environment assessment and medication reviews to reduce use of cardioactive drugs with the potential to cause syncope, and of central nervous system suppressants (Cosman et al., 2014). The most common pharmacological intervention is the use of bisphosphonates, a class of drug which reduces bone resorption through binding to hydroxyapatite in bone and interfering with

osteoclast activity to resorb the tissue (Drake et al., 2008). This medication carries both acceptability issues and risk, however. Especially in earlier agents in the class, there was a requirement to remain upright for 30 minutes after ingestion, and gastrointestinal side-effects were common. The mechanism of action is increasingly being seen to run the risk of atypical femoral fracture, as the microfractures which occur and are remodelled routinely in areas of stress, such as the sub-trochanteric region of the proximal femur, propagate and may lead to atypical femoral fractures in the absence of the osteoclastic activity which is part of the remodelling process (Schilcher et al., 2011). Estimates of the incidence of this problem vary considerably, but a contemporaneous study from the USA reported incidence of 1.78 per 100,000 population in those on bisphosphonates for less than two years, rising to 39.8 per 100,000 once the time on the medication was between six and eight years (Al-Ashqar et al., 2018).

1.3 Management of hip fractures

1.3.1 *The rationale for operative hip fracture care*

The key driver for operative management of hip fractures is the prevention of pre-terminal decline. Failure to recover some form of mobility renders these already vulnerable patients liable to develop venous thromboembolism, chest infections and pressure sores. Problems with personal hygiene, independence and activities of daily living are exponentially greater in a patient with an un-fixed fracture. Pain is also a key indication for surgery; the resultant reduction in autonomic drive confers a physiological benefit and, from a humanitarian perspective, it has become accepted that fixation or replacement may be a palliative procedure with the sole goal of rendering patients more comfortable in their last days. While the drive to modernize hip fracture care has arguably gathered pace in the last decade, Evans argued in 1949 that hip fracture surgery should be considered a surgical emergency, with frailty increasing this urgency, and that pain and immobility were key indications for surgery and so this modernization (Evans, 1949). For the majority of patients, the goal of surgery remains to immediately restore mobility, to permit rapid rehabilitation and to ameliorate the reduction in quality and activities of daily life.

The introduction of standards of care in the UK, detailed in the National Institute for Health and Care Excellence's Clinical Guideline 124 and audited by the National Hip Fracture Database, has centred on rapid assessment by a multidisciplinary team, timely, consultant-delivered or supervised surgery and remobilization from bed the day after surgery. Around this, the package of care includes falls screening, bone health screening, confusion assessment and nutritional care (National Clinical Guideline Centre, 2017; Royal College of Physicians, 2017). These standards have been incentivized by means of a Best Practice Tariff, a payment made to trusts achieving key performance indicators in the clinical guidelines (Oakley et al., 2017). These include timing of surgery, orthogeriatric review and assessment of cognitive capability.

1.32 *History of surgical interventions*

The modern understanding of hip fractures began with Sir Astley Cooper's 'Treatise on fractures and dislocations of the joints', a comprehensive description of a number of clinical cases and anatomical specimens relating to orthopaedic trauma (Cooper, 1822). In this, he described 'Fracture of the cervix femoris' (Figure 1.2) and outlined clearly the distinction between fractures 'external to the capsular ligament' and those within, namely that the former go on to unite and the latter do not.



Figure 1.2 – Fracture of the cervix femoris
(Cooper, 1822)

Smith-Petersen et al. (Smith-Petersen et al., 1931) described internal fixation of intracapsular fractures with a flanged nail in 1931 (Figure 1.3), themselves citing descriptions of previous nails they deemed too bulky for the intramedullary canal and felt caused bone necrosis through pressure, and resultant clinical failure of the construct.

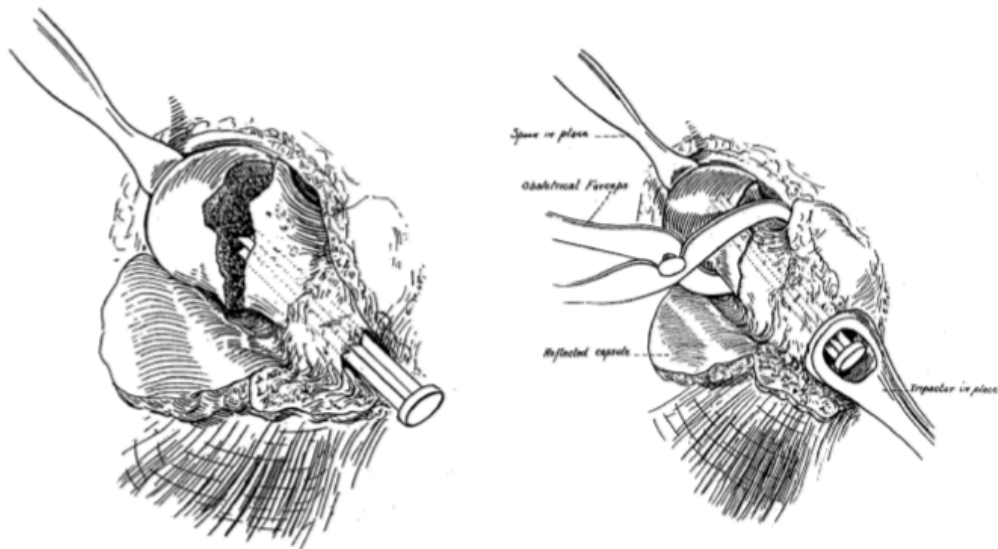


Figure 1.3 – The Smith-Petersen nail
(Smith-Petersen, 1931)

A popular option at that time, while fixation was becoming more commonplace, was augmentation of the construct with prolonged plaster of Paris immobilization. While a surprising number of fractures went on to unite in the context of often wide surgical exposure, the primary benefit of early mobilization was still forgone and so patients remained exposed to the risks of blood clots, chest infections and pressure sores inherent in prolonged immobility (Johansson, 2009).

Hawley patented a plate in 1938 (Figure 1.4), designed to resist bending or breaking, after the American College of Surgeons' Fracture Committee identified the problem as commonplace and needing resolution (Hawley & Padula, 1938). The plate was flanged and, rather than working purely as an onlay implant, was fixed into a channel cut into the bone before being fixed with screws through both the traditional holes at the bone's surface and those through the implanted, flanged region.

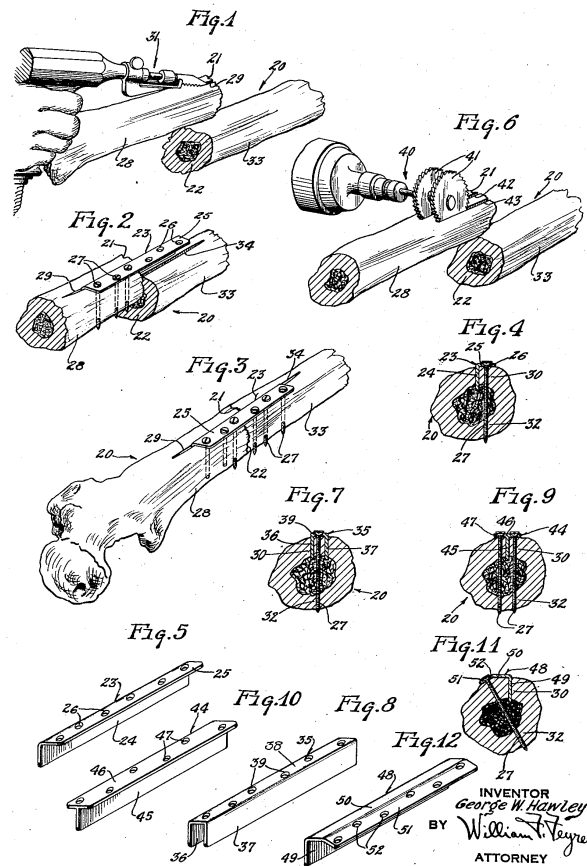


Figure 1.4 – The Hawley plate
(Hawley, 1938)

The Jewett nail plate (Figure 1.5) was the natural evolution of these concepts, offering both fixation of the femoral neck itself with a robust implant and secure fixation in the femoral shaft (Jewett, 1941). He had previously been using both the Smith-Petersen nail and the Hawley plate for dual-fixation, but noted the propensity of the femoral head to rotate about the Smith-Petersen nail. He hypothesized that by combining the two devices, both strength and stability would be gained.

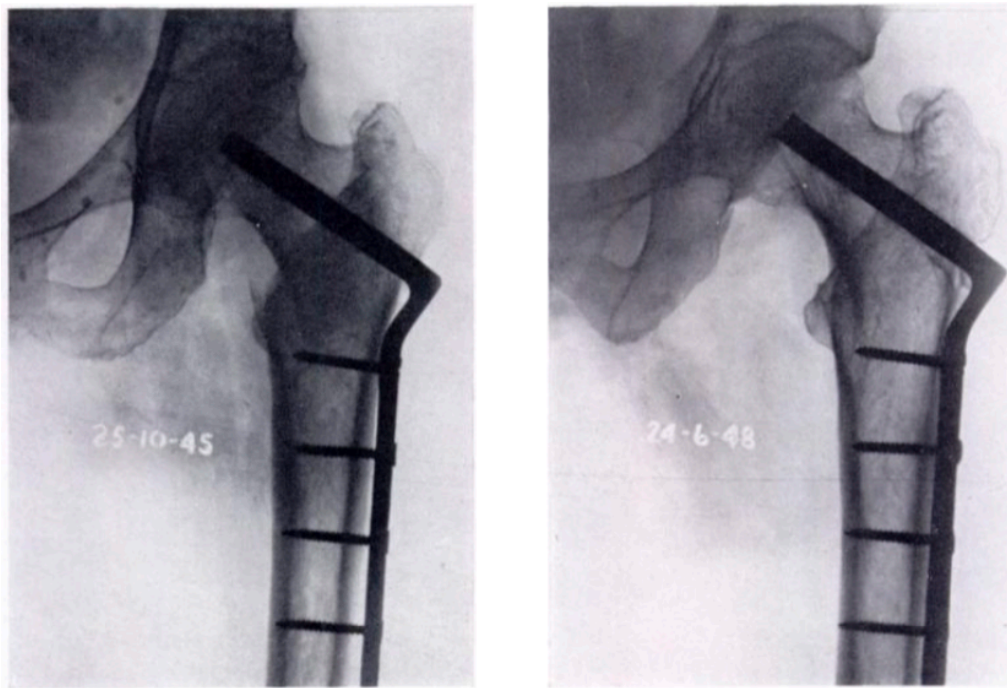


Figure 1.5 – The Jewett nail plate
(Evans, 1949)

The fixed-angle, non-collapsible nature of these devices was controversial in some quarters, and parallel work in Germany by Ernst Pohl led to the patenting of the first sliding hip screw in 1951 (Bartoníček & Rammelt, 2014). This utilized a compression screw across the fracture, able to move freely in the lateral femoral fragment. A barrel plate permitted load to be diverted to the proximal femur whilst maintaining an ability for the screw to slide inside the barrel, thereby allowing the fracture to collapse into maximal compression and hence to maximize chances of healing. A variant of this device was brought into service in the USA by the Richards company as the Richards hip screw, and the first five-year results published in 1964 were very supportive of its use (Clawson, 1964).

British work from some time later, reflecting the delay in uptake of sliding hip screw implants, compared Jewett nail plates and Richards screw plates and found shorter length of stay and better return to mobility with the Richards device, albeit in a small and retrospective study (Heyse-Moore et al., 1983).

The 1970s saw the evolution of the dynamic hip screw in Europe. Although early, small case series reported heterogeneous results, the literature reflects an apparent growth of confidence amongst surgeons, with widening of indications and hence

growth in numbers performed (Oehler & Janka, 1983; Jacobs et al., 1976; Poigenfurst et al., 1983; Wolfgang et al., 1982).

Pohl's work was in evidence once again in the 1980s, when the Gamma nail was developed by Howmedica. It was an evolution of the "Y-nail" patented by Pohl in the 1940s and used for a further 30 years or more within Germany, but never successfully exported. This device evolved rapidly to correct a number of problems such as excessive curvature, a tendency to cause greater trochanteric fractures and to rotate in the absence of a distal locking screw (Halder, 1992).

The principle was adopted and improved by the AO/ASIF foundation and the Proximal Femoral Nail and later Proximal Femoral Nail Antirotation were developed. Most trauma implant manufacturers now offer such a device, with various unique aspects to differentiate offerings from different manufacturers.

There is also an ongoing discussion in the literature about arthroplasty surgery in place of fixation for extracapsular fractures. Whilst this goes against conventional thinking about conserving the native femoral head, there is some logic to this approach. When hip fractures occur in patients with osteoarthritis, extracapsular fractures have been shown to predominate (Dequeker et al., 1993). It may be, therefore, that salvaging a femoral head which is already diseased may offer less functional benefit to the patient than replacing the hip, resulting in a physiological and painless range of motion. The evidence for this is as yet relatively undeveloped, with some small and non-randomized trials reporting heterogeneous results in terms of functional outcome and equivocal results in terms of mortality and morbidity (Duriez et al., 2016; Bonneville et al., 2011). It is important to consider this a concept distinct from arthroplasty for failed fixation, which is far more commonplace and well-reported in the literature (Dix et al., 2018).

1.33 *Current standard of care devices*

The SHS (Figure 1.6) remains in regular use in all units featuring in the NHFD (Royal College of Physicians, 2017). Modern variants tend to be made from stainless steel, with barrel plates featuring a fixed shaft-neck angle usually of 135° but with 130° , 140° , 145° and 150° options available. The plates often offer locking options to improve pull-out resistance in osteoporotic bone, and short-neck barrels are available to ensure collapse and compression remains possible when shorter screw lengths are used in patients with shorter necks. Trochanter stabilizing plates are also available which attach over the barrel plate, with the screws passing through both plates, and grip the greater trochanter, aiming to reduce rates of failure in fractures with comminuted trochanteric patterns.

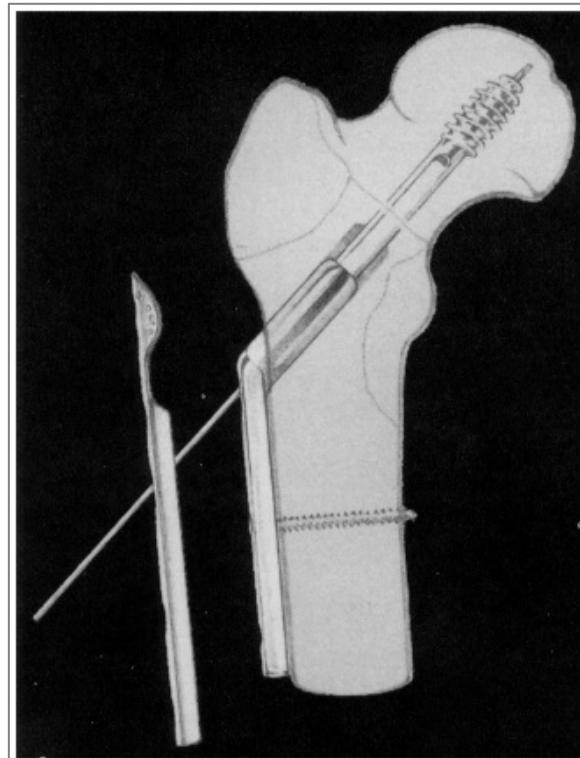
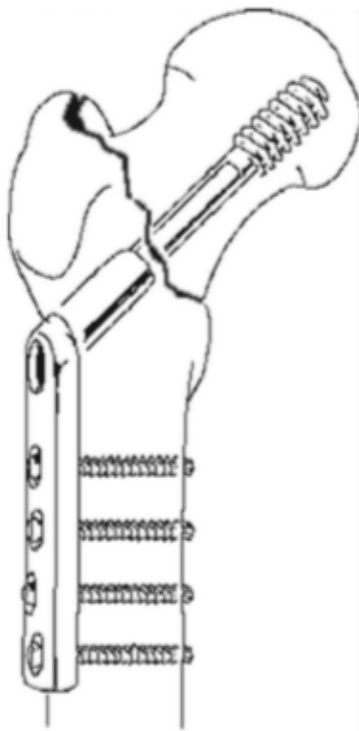


Figure 1.6 – the sliding hip screw (a) in situ and (b) being augmented with a trochanter stabilizing plate

(Schipper et al., 2004; Babst et al., 1998)

Modern intramedullary nails (Figure 1.7) tend to be manufactured from titanium alloy and are available in short (usually around 200 mm) and long (various sizes to

reach the distal physal scar of the femur) variants. The compression screw passes through the nail and in some devices is locked by means of a conical-tip bolt which engages into a channel on the screw (preventing rotation but permitting collapse, while in others it is left entirely unlocked. Distal locking options usually comprise round, static holes and oblong or elliptical dynamic locking holes. If these are used, the locking bolt may be passed through the centre of them, thereby permitting longitudinal compression but preventing rotational movement – most useful in more transversely-oriented fractures. In short nails, distal locking is performed through the introducer jig, whereas in long ones it requires stereotactic surgical skills and the use of an image intensifier.

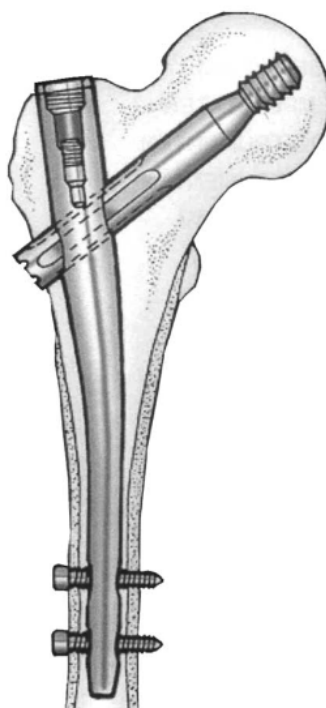


Figure 1.7 – A modern short intramedullary nail
(Schipper et al., 2004)

As usage of these devices has grown, alongside surgeon confidence and competence, there is now a concern that these devices may be being over-used (Page et al., 2016). In the latest versions of the NICE guidance for hip fracture, the indication for use of a nail remains an unstable intertrochanteric or a subtrochanteric (31A3 or 32x) fracture, and an increasing level of surveillance of the use of intramedullary nails (IMNs) in 31A2 fractures has been introduced within the National Hip Fracture Database (Royal College of Physicians, 2017). In addition to the year-on-year trends, there is also a wide variability between centres

in the proportion of trochanteric fractures fixed by SHS. In the 2014 NHFD report, this ranged from 100% of intertrochanteric fractures fixed by SHS at one centre, to 35% at another (Royal College of Physicians, 2014). This continues to be cited as a concern warranting local investigation in the 2017 report (Royal College of Physicians, 2017).

1.4 Current clinical evidence for fixation strategies

1.41 Evidence from existing meta-analysis

The fixation of intertrochanteric fractures has been repeatedly investigated, with a range of evidence from anecdotal to meta-analysis to be found in the literature. Yu's recent meta-analysis of internal fixation options for intertrochanteric fractures provides a clear summary of the evidence at the time of its publication in 2015 (Yu et al., 2015). This meta-analysis included 43 trials, reporting data from 6,911 patients. This evidence has been reviewed a number of times, and this systematic review represents the most recent and rigorous such publication.

Quality of life was reported by EQ5D as higher in Gamma nail than SHS in one trial and similar between Gamma and PFNA nails in another, with the only other two trials reporting quality of life measures lacking pre-injury data. While capturing pre-injury quality of life scores post-injury is complex, there are some acceptable strategies to do so. Overall, quality of life scores were lower at the end of the reporting period than at baseline. Functional scores, although heterogeneous between studies, found the Gamma nail superior to SHS but inferior to the PFNA. The SHS was, in turn, superior to the intramedullary hip screw (IMHS). No bias-free data was available on mortality, and the sample sizes and incidence of death were sufficiently small that none of the included studies were deemed likely to offer statistically robust evidence.

An overall incidence of 141 in 3,692 patients (3.5%) was seen of cut-out of the device from the femoral head, where the head progressively cavitates around the device until the screw perforates the head and it effectively falls off. This complication is important as it mandates re-operation in all but immediately pre-terminal patients. The Gamma nail showed an increased rate of cut-out, at 43 in 802 fractures *versus* 23 in 830 fractures fixed by SHS. Non-union was reported in 29 trials, comprising 3,795 patients, in whom 90 (2.37%) events were reported. The only statistically significant comparison between implants was a superiority of SHS over IMHS, with all other data hampered once more by small sample sizes.

Re-operation data was similarly problematic, but an increased incidence of re-operation with Gamma nails (53 in 907 patients) over SHS (35 in 939 patients)

was statistically significant. This conclusion is logical when considered in the context of the cut-out rate shown with the two devices.

Data on intra-operative fracture was structured in such a way as to include those fractures recognized during the follow-up period but deemed iatrogenic. Data was pooled on 2,661 patients and showed a significantly increased risk of fracture with Gamma nail over SHS (22 *versus* 5 in 861 patients) and with Gamma nail over PFNA (19 *versus* 5 in 125). The only statistically significant comparison of devices when considering later fractures found an increased risk with the Gamma nail over the SHS (18 of 703 *versus* 2 of 704).

No significant differences between devices were reported in terms of wound infection. Data on embolic events were available in 2,655 patients and, with 88 events in total, the only significantly increased risk was seen in SHS (11 in 87) over percutaneous compression plate (PCCP, 4 in 92).

Operative time data varied widely. The PFNA was found to have a lower operative time than the Gamma nail, Less Invasive Stabilization System (LISS) and PCCP. It did not, however, exhibit a time advantage over the SHS.

Intra-operative blood loss was reported in 19 trials and, overall, the SHS caused more blood loss than the other, heterogeneous group of implants with which it was compared. In the context of the minimally invasive technique by which IMN are implanted, however, these results should be interpreted with caution as a true marker of blood loss in patients undergoing IMN may in fact be postoperative haemoglobin measurements, or the need for transfusion.

Hospital stay was reported in 22 trials, with the only significant comparison being an increased length of stay in PFN over Gamma nails.

1.42 Recent updates to the evidence

To identify further trials of interest since this meta-analysis, Yu's search strategy (see Appendix 1) was re-deployed in the PubMed, EMBASE and CENTRAL databases with date limits for the period between May 2015 and March 2018. This yielded 1,142 citations in PubMed, 2,599 in EMBASE and 111 in CENTRAL. After scrutiny, 9 PubMed citations were identified as eligible for inclusion, 14 from EMBASE and 11 from CENTRAL. Once duplicates were excluded, 13 papers remained. These are summarized in table 1.1.

Table 1.1 – Recent RCTs relating to the fixation of intertrochanteric fractures

Study	Population	Intervention	Control	Outcome	Findings
Reindl (2015)	204 pts ≥55yrs	IMN	SHS	LEM, TUG, FIM, 2MW Fracture movement, HO, failure Complications	No clinical superiority Improved radiological markers
Sanders (2017)	249 pts ≥55 yrs	InterTAN IMN	SHS	FIM, TUG Femoral shortening Mortality	No clinical superiority except in those walking >150m pre-injury
Cai (2016)	198 pts ≥65 yrs	IMN	SHS	Blood loss, electrolytes FRS Radiological union Cut-out	Increased blood loss in IMN Worse electrolytes in IMN Hypoproteinaemia in IMN
Parker (2017)	400 adult patients	Targon PFT IMN	SHS	Reduction in mobility scale Pain scale	Improved mobility at 3, 6, 9 months, not present at 1 yr
Zehir (2014)	198 adult patients with unstable fractures	PFNA IMN	SHS	Operative time, fluoroscopy, TA, cut- out, recovery of and independent walking	No difference
Hopp (2016)	78 pts ≥ 65 yrs with unstable fractures	InterTAN IMN	Gamma IMN	Harris hip score Complications	No difference
Andreani (2015)	81 pts with 31A1/A2 fractures	ENDOVIS IMN	Gamma IMN	Parker-Palmer mobility score Barthel index	No difference

				EQ5D Operative time, transfusion requirement	
Özkayin (2015)	54 pts \geq 75 yrs	PFN	Hemiarthroplasty	Harris Hip Score (HHS) Complications	Better early HHS with hemiarthroplasty, PFN superior by 12 months
Desteli (2015)	86 pts \geq 60 yrs	PFN	Cementless bipolar hemiarthroplasty	EQ5D	PFN superior mobility and social function scores at 2 yrs
Berger-Groch (2016)	104 skeletally mature pts	InterTAN	Gamma	LOS SF36, Harris hip score Varus collapse, non- or mal-union, TAD	InterTAN superior functional outcomes at 6 months, no difference at 5 years
Shin (2017)	353 pts \geq 65 yrs	Zimmer Natural Nail	PFNA	Harris hip score Operative & fluoroscopy time Lateral hip pain Walking ability Cut-out	ZNN more lateral hip pain than PFNA
Li (2015)	70 pts \geq 65 yrs with 31A1/A2 fractures	PFNA with distal locking	PFNA without distal locking	Operative time, blood loss, fluoroscopy time Wheelchair use Recovery of walking	Unlocked shorter operative time, less blood loss. No functional difference.
Griffin (2016)	100 pts \geq 60 yrs with 31A2/A3 fractures	X-Bolt dynamic plating system	SHS	EQ5D	No difference

There has been a particular focus in recent trials on the 31A2 subgroup, those fractures with some comminution but without significant extension below the lesser trochanter. It is in this subgroup that an increased rate of nailing seems to be evolving, but in two adequately powered trials designed to detect the minimum clinically important difference in their respective outcome measures, no major advantages were reported.

Reindl et al. randomized 204 patients aged 55 or over to either SHS or short IMN (Reindl, Harvey, Berry, Rahme, et al., 2015). The primary outcome measure was the Lower Extremity Measure (LEM), with the Timed Up and Go (TUG) test, Functional Independence Measure (FIM) and 2-minute walk assessed as secondary outcomes. Tip-apex distance and femoral neck shortening were assessed as radiographic outcome measures. Fixation with IMN conferred no functional benefit, and although better radiological outcomes were seen in this group, this was a secondary outcome measure and was not associated with a clinical benefit.

Sanders et al. conducted a multi-centre randomized controlled trial (RCT) comparing the InterTAN IMN with SHS, including 249 patients aged 55 and over (Sanders et al., 2017). The study reported FIM, TUG, femoral shortening, complications and mortality. Once more, the IMN offered no significant clinical benefit across all patients. When confined to a subgroup analysis of those patients reporting a pre-injury ability to walk >150m, improved scores were seen, suggesting the IMN was a better choice in pre-morbidly active patients but this subgroup analysis was secondary and at high risk of bias.

Cai et al. compared overt and hidden blood loss between intra- and extramedullary fixation and found that chest infection, hypoproteinaemia, electrolyte imbalance and overt and hidden blood loss were all lower in the SHS group (Cai et al., 2016). On this basis they advocated a preference for SHS over IMN in stable intertrochanteric fracture configurations.

Parker et al. reported an RCT of 400 patients with 31A1, A2 or A3 fractures randomized to SHS or IMN and found that the only difference to be an improved

regain of mobility at one year {Parker:hv} . This is slightly less clear when one considers the IMN implant used changed throughout the trial, a limitation acknowledged by the author. The combination of this evidence and that from Sanders' trial may suggest, however, that a better understanding of the mobility factors in IMN and SHS fixation would be beneficial.

Zehir et al. reported better recovery of independent walking in a RCT comparing PFNA with SHS, but equal incidence of complications (Zehir et al., 2014). The methodology does not, however, describe any sample size calculations and the groups are factorially smaller than those of other, well-powered trials, raising the suspicion that they may be convenience samples.

Hopp et al. conducted a similarly small RCT comparing the third generation Gamma nail with the InterTAN IMN and found no difference in Harris Hip Score or complications (Hopp et al., 2016). The participating surgeons rated the InterTAN as more cumbersome, but this is both subjective and predictable on the early phase of any learning curve. Patient outcomes were not affected by this and so there is no suggestion to favour one device over the other.

Andreani et al. compared the novel ENDOVIS nail with the contemporary Gamma nail, again with small sample sizes, and found no differences in outcome over wide range of functional scores and perioperative measures (Andreani et al., 2015).

Özkayin and colleagues randomized patients to either PFN or hemiarthroplasty in a small but randomized controlled trial, recruiting 54 patients in total (Ozkayin et al., 2015). The Harris Hip Score initially favoured hemiarthroplasty but, by 12 months post-operatively, those undergoing fixation by PFN had superior scores. It is not explained within the article why the study was testing a strategy of replacement against fixation for these fractures – aside from the very small sample sizes, it is hard to see credibility in equipoise between these approaches. Desteli et al. conducted a similar study, also based in Turkey, finding improved social function and mobility scores at 2 years in patients who had been treated with the PFNA over those who had been treated with cementless bipolar hemiarthroplasty for unstable intertrochanteric fractures (Desteli et al., 2015). It is notable that this paper does, in its introduction, state that there is no consensus on fixation versus

arthroplasty, suggesting that there may be more regional variation in approaches than is generally recognized.

Berger-Groch et al.'s trial comparing the Gamma 3 to the InterTAN nail found limited evidence of superiority of the InterTAN – in addition to a shorter hospital stay, an early benefit of superior pain and functional scores at 6 months was not perpetuated to the end of follow-up at 5 years (Berger-Groch et al., 2016).

Shin et al. compared 2 short nails, the modified PFNA (PFNA II) and Zimmer Natural Nail (Shin et al., 2017). Once more, no differences were seen in outcome, and the only differences seen were in operative and fluoroscopy time. Again, this is a trend to be anticipated when a surgeon familiar with one nail is given an experimental device to implant, due to subtle differences in instrumentation and device-specific technique.

Li et al. sought to assess the benefit of distal locking, by comparing locked *versus* unlocked nails in pertrochanteric fracture patterns (Li et al., 2015). Albeit a small study, it showed significant differences in operative time and blood loss between groups, with no clinical or functional impact of omitting distal locking.

Griffin et al. reported the outcome of a feasibility study to inform the full RCT comparing the X-Bolt dynamic hip plating system with SHS, showing no differences in functional outcome between implants (Griffin et al., 2016). As this was a feasibility study of 100 patients, showing the definitive RCT would need to recruit 964 patients to achieve adequate power, these results should be interpreted in this context.

Overall, since Yu's meta-analysis, there has been no change in the position that a universal benefit has yet to be demonstrated from IMN over SHS in less stable intertrochanteric fractures.

Summary of key clinical points in the literature

Hip fractures are painful, debilitating and functionally challenging for patients and expensive in terms of finance and healthcare resources for the nation. They tend to occur in older patients with reduced bone mineral density due to osteoporosis.

Hip fractures may occur within or outside the capsule of the hip joint, giving rise to intra- or extracapsular fractures, respectively. This is key, as it predicts whether the blood supply to the femoral head has been irretrievably impaired, and hence whether fixation or replacement is the most appropriate strategy.

The treatment strategy for extracapsular fractures is predicated on fixation. This fixation may be with sliding hip screw or intramedullary devices. Except in the very unstable 31A3 fractures, there is no evidence to support a benefit to intramedullary over sliding hip screw fixation. This must not be confused with the principles of treatment of subtrochanteric fractures of the femur, where intramedullary nailing is the gold standard (National Clinical Guideline Centre, 2017).

Chapter 2: Biomechanics

2.1 Biomechanics of the native hip

The hip is a highly congruent ball and socket joint, characterised by 3 degrees of freedom and permitting movement primarily by rotation in three axes.

Figure 2.1 shows forces acting through the hip. In single-leg stance, compressive force (R) acts across the hip and through the centre of the femoral head, but importantly not in the axis of the femoral neck. The muscular forces required to keep the pelvis level, provided by the abductors, act in the line M to the point of gluteal insertion on the greater trochanter. Total body weight, less that of the weight-bearing leg (K), acts through a point just lateral to the contralateral side of the symphysis pubis. The lever arm of the abductors (b), the distance between the greater trochanter and the centre of rotation of the femoral head) is around one third of that between the centre of the femoral head and the contralateral side of the symphysis pubis (a), the point through which body weight less that of the stance leg (K) acts.

In a level pelvis, therefore, $M \times b = K \times a$. A key clinical implication of this is that any change in b, such as by excessive collapse of a fracture during healing, means that either the work done by the abductors must increase, tiring the patient, or the pelvis cannot remain level, and the patient acquires a Trendelenburg gait.

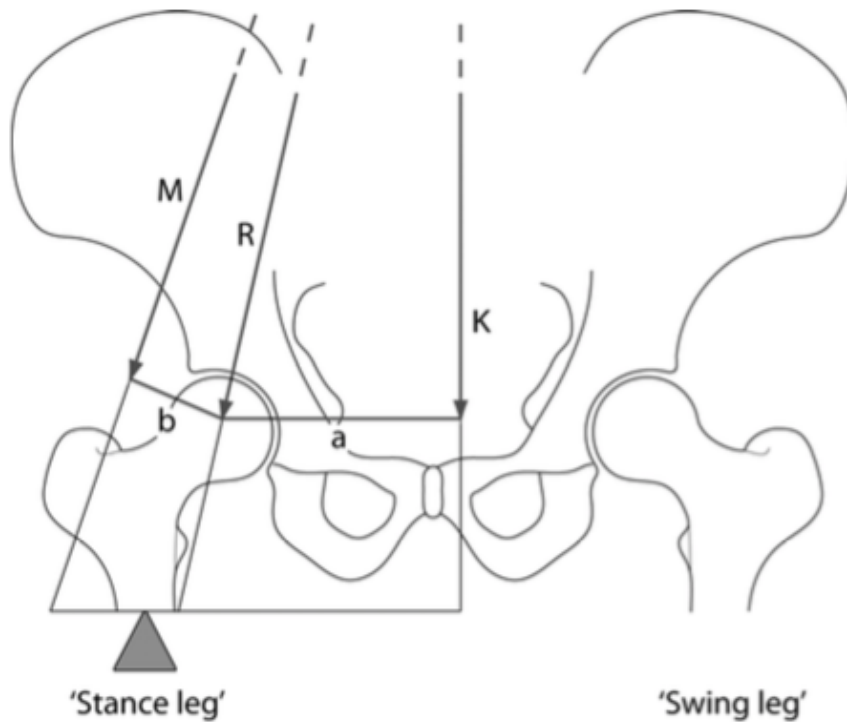


Figure 2.1 – Forces acting through the hip in stance
(Byrne et al., 2010)

These calculations pertain to the joint in a static state, but in fact the joint is subject to forces far in excess of total body weight during everyday activities. The loads acting on the hip have been quantified in a number of theoretical approaches, but also by means of mechanotransduction studies (Seireg & Arvikar, 1975). Bergmann et al. used an instrumented variant of a clinically-proven total hip arthroplasty prosthesis, equipped with strain gauges and powered by induction from outside the hip, to measure forces acting on the hip joint during common movements (Bergmann et al., 2016). This builds on work from the same team, using earlier versions of the technology and smaller cohorts (Bergmann et al., 2001; 2010). Cumulatively, these and other publications have now been aggregated at the OrthoLoad website, an online resource of loading data for biomechanics researchers (Bergmann et al., 2017).

A summary of these forces is reproduced in table 2.1.

Activity Table 2.1 – Forces and moments acting through a total hip replacement implant in a live patient during activities of daily living.

x, y, z = axes of femur coordinate system. x = parallel to posterior contour of condyles. P1 = intersection of neck axis and femoral midline. P2 = middle of intercondylar notch. z = straight femur axis between P1 and P2. Force components F_x , F_y and F_z act in directions x, y and z. Moment components M_x , M_y and M_z turn clockwise around x, y and z. (Bergmann et al., 2016)

		Cycling	Sitting	Standing	Knee bending	Walking	Stance	Stairs up	Stairs down	Jogging
Force (N)	F_x	323/420	442/1002	472/1578	428/1202	265/837	347/956	379/1164	329/1085	311/1241
	F_y	-357/-58	-541/-88	-612/70	-455/-52	-536/24	-203/-50	-1067/-68	-855/-78	-1222/-21
	F_z	-392/-1311	-728/-2709	-753/-3480	-1069/-2882	-235/-2709	-644/-3198	-558/-3876	-405/-3662	-229/-4519
Moment (Nm)	M_x	-0.60/1.11	-2.03/1.57	-1.11/2.61	-1.97/2.39	-1.28/1.47	-1.33/1.41	-1.21/1.82	-2.06/1.19	-1.44/0.97
	M_y	-0.22/0.19	0.00/0.77	-0.70/0.78	0.31/0.93	-0.37/1.21	-0.71/1.71	-0.51/1.24	-0.88/1.50	-0.49/1.07
	M_z	-0.25/0.25	-0.57/0.34	-0.22/0.70	-0.63/0.43	-0.50/0.63	-0.55/0.93	-0.44/0.44	-1.04/0.30	-0.72/0.13

2.2 Biomechanics of hip fractures and their fixation

A key principle in the fixation of hip fractures is the balance of load bearing and load sharing. The final construct should permit the forces acting through the hip to be borne in part through reduced, stable fracture fragments. Implants which bear disproportionate load are prone to failure through fatigue, and inadequately loaded bone risks not setting the right biomechanical environment for bony healing (Bucholz et al., 1987; Augat et al., 2004).

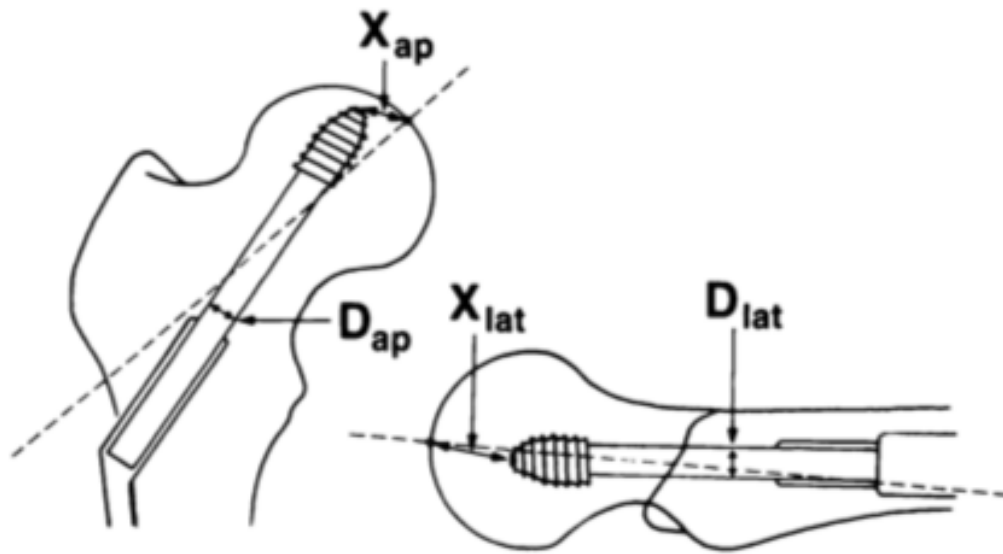
An important element contributing to the stability of a trochanteric fracture is the posteromedial cortex. This thickened bone conducts load to surrounding regions, acts as a radiographically visible landmark to the surgeon when assessing reduction of a fracture and, if intact, provides stability. Comminution here confers relative instability and its recognition is essential when planning surgery. The lateral wall of the greater trochanter is also important especially when preventing excessive medialization of the femur. Especially in already comminuted fractures, it is relatively easy to cause fracture of the lateral wall when reaming for a sliding hip screw (SHS) and so vigilance and anticipation is required (Tawari et al., 2015).

Gundle et al. demonstrated the importance of permitting adequate sliding in the SHS, finding in their prospective study of 100 consecutive patients undergoing SHS for unstable intertrochanteric fractures that those fixed in such a way that the screw was able to slide less than 10 mm were those most at risk of failure (Gundle et al., 1995). The fractures in this study population needed to collapse by more than this distance to create good apposition of the head and neck fragment onto the buttress of the femoral fragment – by preventing this, the load was not shared between femoral neck and SHS screw, resulting in an overloaded implant and a fracture site without adequate compressive contact to stimulate union. Excessive collapse, by contrast, shortens the limb and the lever arm of the abductors, risking a significant Trendelenburg gait disturbance. This has been shown in a number of studies to have a significant association with patient reported physical function outcomes as measured by the Short Form 36 (SF36) questionnaire (Zlowodzki, Brink, et al., 2008; Zlowodzki, Ayieni, et al., 2008). Intramedullary devices offer a buttress against collapse without reliance on the lateral trochanteric cortex, as a collapsing head and neck fragment will encounter the nail which, if correctly used,

will be sharing load along its length and hence relatively reducing the forces exerted on the proximal lateral cortex.

The divergence between the axis of the femoral neck and that through which body weight acts is important in hip fractures. In any fracture, it is ideal to compress it with a force perpendicular to it (Ruedi & Murphy, 2000). Eberle et al.'s finite element study of implant load in an intramedullary nail fixing different fracture configurations showed that significantly more interfragmentary motion was seen in a pertrochanteric fracture, with a force vector not perpendicular to it, than a subtrochanteric one with an almost perfectly perpendicular vector (Eberle et al., 2009).

One of the most pivotal studies in developing understanding of the SHS and its modes of failure was that reported by Baumgaertner et al. (Baumgaertner et al., 1995). This retrospective study introduced a means of assessing and describing the position of the compression screw in the femoral head, by measuring the distance from the tip of the screw to the apex of the femoral head in the AP and lateral views – the tip-apex distance (TAD) (see Figure 2.2a). Patients were grouped by failure – those in whom fixation failed had a mean TAD of 38 mm, compared with 24 mm in those with fixation that did not fail. This has led to a near-universal adoption of a TAD of 25 mm or less as a gold standard when fixing trochanteric hip fractures. It also noted that the lowest incidence of cut-out occurred when screws were placed centrally in the head in both views, with further enumeration of the cut-outs occurring from each zone of the femoral head (Figure 2.2b).



$$\text{TAD} = \left(X_{\text{ap}} \times \frac{D_{\text{true}}}{D_{\text{ap}}} \right) + \left(X_{\text{lat}} \times \frac{D_{\text{true}}}{D_{\text{lat}}} \right)$$

Figure 2.2 – (a) Calculation of the tip-apex distance (b) Total screws placed and number cutting out, by zone of femoral head .

Dtrue – known diameter of compression screw, used to calibrate measurements.

(Baumgaertner et al., 1995)

2.3 Design considerations for fixation devices

2.3.1 Intramedullary nails

Buchholz's finite element analysis study based on seven patients who had experienced periprosthetic fractures distal to the tip of a long femoral nail found that the femur would have had to regain 50% of its stiffness through healing before the nail could safely be loaded without expectation of fatigue failure (Buchholz et al., 1987). The analysis also identified, however, that the fracture was within 5 cm of the distal locking screws in all the patients. This early work into intramedullary nailing highlighted a problem but arguably offered the wrong solution, as there now exists a clear imperative to use the devices as part of a patient's rapid return to bearing weight (Moran et al., 1990; Brumback et al., 1999). It may have been more pertinent to highlight that the choice of a shorter nail with little working length distal to the fracture conferred high risk of failure. The authors' other recommendation was far more in keeping with today's principles, however, and was that larger diameters of nails should be used. By increasing the diameter, more of the nail's surface area is in contact with the femur and hence more load is shared. At its extremes, under-sizing of a nail can lead to the situation where most dissipation of load is through the distal locking screws, significantly raising the likelihood of failure through periprosthetic fracture (Erduran et al., 2011).

Schneider et al. performed an *in vivo* study of forces acting on a femoral nail, using a telemetry-equipped nail which reported torsional and axial strain (Schneider et al., 2001). This showed a substantial decline in loading of the nail once the fracture consolidated, with around a 50% reduction seen following consolidation.

Other characteristics may vary between nails. Especially in earlier generation nails, a substantial difference in stiffness was seen between proximal, middle and distal parts (Russell et al., 1991). This, therefore, obliges knowledge of the particular nailing system and to ensure the length and performance of the implant are appropriate to the fracture being managed, to ensure the fracture is being bridged by an adequately robust construct.

The metaphyseal and proximal parts of modern hip nails must be of adequate diameter to permit passage of the compression screw through them, with

sufficient remaining nail around the hole to cope with the loads placed upon it and to house any locking mechanism the device may employ. This in turn mandates caution around sizing of the nail, as an excessive diameter can result in the very stiff portion of the nail propagating or comminuting the fracture substantially.

The length of nail may be short or long; the short variant has a distal locking screw facilitated by a targetting jig, while the long variant requires freehand distal locking with stereotactic radiographic techniques. Historically, there have been some concerns over the use of short nails due to the risk of a stress riser at the distal tip of the nail, but most documented instances of this complication pertained specifically to the first generation of modern cephalomedullary devices, in particular the Gamma nail (Bridle et al., 1991; Williams & B. C. Parker, 1992; Leung et al., 1992). More recent studies have shown no difference in peri-prosthetic fracture between short and long nails, and there exists an overall paucity of robust, prospectively-gathered evidence (Boone et al., 2013; Kanakaris et al., 2015). This is discussed and illustrated further in Chapter 4.

2.32 *Rotational stability*

There exists an inherent variability in how rotationally stable a construct created by fixation of a hip fracture is. One important variable is the fracture location relative to the head – a subcapital fracture (less commonly fixed as the blood supply is likely to be compromised, leading to the need to carry out a replacement) tends to have less inherent stability due to a small surface area for interdigitation and a fracture pattern which does not lend itself to the keying together of principal fracture fragments. At the other end of the scale, an intertrochanteric fracture has a large surface area to interdigitate, and may feature some form of apex which, when reduced, prevents the fragments rotating during either surgery or walking (Ruecker & Rueger, 2014).

Two forms of rotation are possible; in unlocked implants, the head/neck and screw fragment may rotate as one, with the screw rotating within the barrel of the implant, whereas when this screw is locked, the rotation must be of the fracture fragment about a locked screw.

While there have not been large clinical studies of the effect of rotational stability, a retrospective study of 115 patients with SHS fixation found that 14 of them had radiographic evidence of rotation of the head and neck fragment, conferring a significantly higher risk of non-union, cut-out and varus angulation. The incidence of this rotation was higher in the group with more complex fracture patterns (Lustenberger et al., 1995). A more recent radiostereometric analysis (RSA) study showed that in trochanteric fractures, rotation of the fragments relative to each other occurred up to 4 months post-operatively and found left-sided fractures to be inherently more unstable than right-sided ones. This is postulated to be due to the compression induced between neck fragment and buttress when screwing an implant in clockwise on the right, whereas no such buttress is encountered when the same clockwise torque is induced on the left (van Embden et al., 2015).

Lenich et al. performed a biomechanical experiment to quantify the resistance to rotation of two implants, the PFNA blade and the SHS screw (Lenich et al., 2011). This demonstrated superior resistance in the PFNA blade. Importantly, the argument made for the importance of rotational stability was that, should a fracture be fixed with the implant off-centre, a rotational motion could have a

cavitating effect within the femoral head and neck, thereby creating a void rendering the construct prone to cut-out.

Both Aguado-Maestro and Massoud reported successful strategies to mitigate the risk of rotational instability in basicervical fractures by supplementing their chosen fixation with additional derotation screws (Figure 2.3) – although neither was a large or methodologically robust trial, the approach has face validity and suggests that addressing the rotational element of these fractures is important (Maestro, 2013; Massoud, 2009).



Figure 2.3 – Augmentation of a SHS with derotation screw
(Massoud, 2009)

Rotational stability is often discussed when debating fixation with IMN *versus* SHS, but the evidence to support an inherently increased stability in IMNs is lacking, not least because the overall concept of rotational stability has not been investigated in great depth. What an IMN does typically offer is the ability to lock the compression screw within the barrel of the nail, thereby guarding against rotation of the head and screw as a single construct. This contrasts with a SHS, where the screw can freely rotate within the barrel plate.

Intramedullary devices also offer the potential dual-screw fixation, such as that used in the InterTAN device (Figure 2.4), whereby an initial fixation is made with one screw, then a second used to engage and lateralize the first, and hence the femoral head, relative to the IMN. The resulting fracture is then compressed by a figure-of-8 shaped construct of two screws, theoretically more resistant to

rotation about its axis. Santoni et al. demonstrated this increased stability when comparing a conventional single-screw fixation with the InterTAN in a cadaveric hemipelvis model, with significantly less rotation seen in the InterTAN model (Santoni et al., 2016).



Figure 2.4 – The InterTAN intramedullary nail with dual-screw system.
(Santoni et al., 2016).

Ma et al.'s meta-analysis of studies comparing the InterTAN with PFNA or Gamma nails noted an decreased rate of cut-out and periprosthetic fracture but, interestingly, this was not reflected in the primary outcome measure of the study,

the Harris Hip Score (HHS), which was not significantly different between groups (Ma et al., 2017). This should not be confused with fixation by two separate screws; in this scenario, there exists a risk of the “Z effect phenomenon”, where fixation with two, separate compression screws results in lateral migration of the inferior screw and medial migration of the superior screw (Strauss et al., 2007).

2.33 *Torsional stability*

Rotational stability has traditionally been taken to refer to rotation in the axis of the femoral neck, i.e. around the compression screw. The other rotational force important in hip fracture biomechanics is that around the longitudinal axis of the femur (Figure 2.5). When using an IMN, this axis should be similar to that of the nail itself if it has been inserted correctly.

Bergmann et al. demonstrated increased contact pressures in the hip during stair climbing, and also quantified the forces and moments acting in the anteroposterior (AP) plane during this motion, rising to stand and sitting down (Bergmann et al., 2001; 2016).

When considering that these actions involve AP loading of a flexed hip, it can be seen that this is effectively a vertically-oriented pelvis pushing on a horizontally-oriented femur.

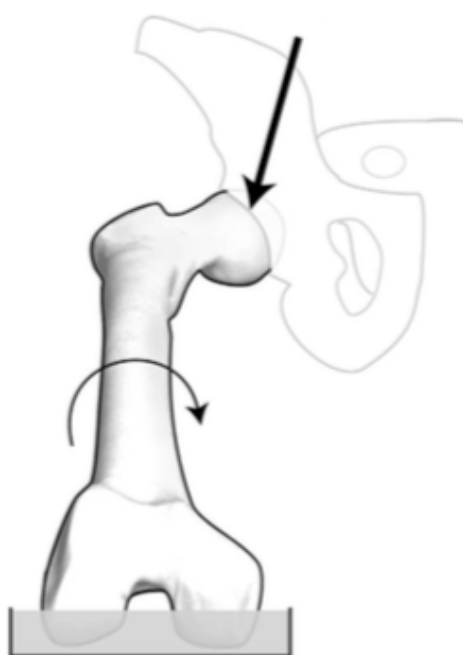


Figure 2.5 – The biomechanics of torsional stability
(Basso et al., 2012)

The implication of this loading has not been widely studied. Most biomechanical experiments load the femur axially in its vertical orientation, which reflects the stance phase of gait reasonably well, but does not address these movements and the increased forces they can entail. Blair et al. compared a trio of cannulated

screws, a SHS and a SHS with derotation screw in a cadaveric biomechanical experiment and included a torsional element to the study (Blair et al., 1994). It did not include an intramedullary device and found no difference between implants but, importantly, is one of few experiments to investigate torsion.

2.34 The X-Bolt system

Recently, a new device has been brought to market with the specific aim of reducing the incidence of cut-out, in part by improving this rotational stability. The X-Bolt is an expanding, cruciform implant which is deployed within the femoral head once the surrounding cancellous bone has been compacted. By means of a threaded central portion, the wings of the device are shortened and hence begin to fold at the point machined to be their apex, visible in the diagram below. This device acts in lieu of a compression screw and hence may be deployed through a barrel plate in the manner of a conventional SHS, a mini-plate or an IMN. Figure 2.6 shows this device in its SHS configuration.



Figure 2.6 – The X-Bolt hip fracture fixation device

The design has been shown to have superior rotational stability to both screws and the helical blade design of the PFNA in an osteoporotic bone substitute, although this finding has yet to be reproduced *in vivo* (Gosiewski et al., 2017).

Biomechanical testing of pushout resistance, the force required to advance the implant through the femoral head, showed both a lower gradient of displacement against force when compared with SHS and SHS blade devices and also a plateau around 400 N (O'Neill et al., 2013). In practice, this suggests both that

displacement is slow, and that its peak value may be relatively lower than with other implants. The device is currently in the experimental arm of a major clinical trial (<https://www.ndorms.ox.ac.uk/clinical-trials/current-trials-and-studies/white>), having previously been found in a feasibility trial to confer similar patient reported outcomes and rates of complication to the SHS (Griffin et al., 2016) .

2.35 *Metallurgy*

The early choice of material for surgical devices in general was stainless steel, which had good availability, a relatively reliable and reproducible manufacturing process and tended to be affordable in those healthcare economies using such devices (Disegi & Eschbach, 2000). Whilst the original discovery of the low-carbon alloy was made in 1904, it was in 1926 that the 18Cr-8Ni alloy used in medical devices was described. By varying the degree of annealing of the metal, the ductility can be altered at the expense of tensile strength – at one end of this spectrum is the highly ductile but relatively weak wire used for cerclage, for example, at the other, the much more bending-resistant Kirschner wire, designed to be driven into bone a number of times with a relatively long working length between driver chuck and cortical surface.

Medical titanium alloys have become increasingly popular over recent decades, with a common alloy being Ti-6Al-4V. An improved tissue compatibility has been cited as a key driver for this; in addition to biomechanical advantages, a reduced tendency to deep infection has led to a marked preference for titanium alloys over stainless steel (K. Wang, 1996; Arens et al., 1996).

The stiffness of the metal used in these implants is important. The tendency of a metal to deform on an axis when a force is applied to that axis is termed Young's modulus, or the modulus of elasticity, and may also be understood as the ratio of strain to stress. The Young's modulus of any implant must be one sufficient to prevent unwanted deformation when loaded, but also avoid the stress shielding associated with a steep gradient in Young's modulus between implant and bone. Stress shielding is the resorption of bone caused by reduced loading, the force being transmitted preferentially by the implant, and characterized by a radiological appearance of osteopenia and risking periprosthetic fracture (Wilson & Hench, 2012). The Young's modulus of medical-grade stainless steel is 186 GPa, contrasting with around 120 GPa for Ti-6Al-4V titanium and 105 GPa for Ti-12Mo-3Nb, one of the newer β alloys (Kuroda et al., 1998; Gabriel et al., 2012).

The choice of metal alloy clearly, therefore, has implications for how an implant can be expected to perform. The material property of stiffness is one which will be the focus of one experiment in this study.

2.4 Methodology in biomechanical experiments

2.4.1 *Bone and substitutes*

A number of studies of hip fracture biomechanics have used cadaveric specimens (Haynes et al., 1997; Curtis et al., 1994). This offers good generalizability to human patients, but can offer some challenges in terms of uniform material properties, given the individual differences in bone mineral density one would expect to see across a population. Furthermore, the restrictions of the Anatomy Act and the regulatory framework of the Human Tissue Authority make it challenging to use these materials, with dedicated laboratories and instrumentation usually required on the same site licensed under the Act.

The use of synthetic bone substitute materials for biomechanical testing has become increasingly commonplace as designs have evolved to provide more and more fidelity to human bone. A number of biomechanical characterizations have shown that, with each generation, the synthetic bones have performed within the range seen in human bone for axial loading, torsion, 4-point bending and viscoelastic deformation (Cristofolini et al., 1996; Heiner, 2008; Heiner & Brown, 2001).

Animal models have also been shown to have acceptable characteristics to be used in experimental models. Recent work on bovine tibia has validated an osteoporotic long bone model, using acid degradation to produce specimens with similar BMD to those in osteoporotic human long bones, and exhibiting similar resistance to screw pull-out (Fletcher et al., 2017).

O'Neill et al. showed that polyurethane foam of a density of 0.16 g/cm^3 had similar performance to the cancellous bone within cadaveric femoral heads on force-displacement curves, demonstrating not only that an intact model had similar properties, but also that instrumented substitutes could be expected to behave similarly to their cadaveric counterparts (O'Neill et al., 2012). This effectively validated the work of Patel et al, who had performed biomechanical testing on this foam and found it to be compatible with the known properties of human cancellous bone, but not tested the two side by side (Patel et al., 2008).

2.42 *Simulating physiological forces at the hip*

Simulation of some part of the physiological loads acting on the hip are required to obtain generalizable information from an experiment. Many studies have simulated gait principally using an axial compressive force applied to the superior pole of the femoral head, orienting the specimen in such a way that the load is applied with a vector as close to physiological as possible in the coronal plane (Haynes et al., 1997; Sommers et al., 2004; Curtis et al., 1994). A biaxial rocking motion has also been described, applying this axial force via an obliquely-oriented contact surface, thus exerting axial and varus force on the femoral head model (Ehmke et al., 2005).

Rotational stability has been assessed primarily either by load to failure, the force required to induce rotation of the femoral head and neck, or by in vivo studies observing the rotation seen after physiological loading (Lustenberger et al., 1995; van Embden et al., 2015; Gosiewski et al., 2017). The only experiment in the literature using a simulated physiological load was that by Lenich et al., but this again used a forced axial rotation to induce the load (Lenich et al., 2011).

2.43 Computer-based modelling

The evolution of finite element analysis (FEA) has permitted the modelling of complex constructs and scenarios with an almost infinite number of permutations. This is important as it permits vastly more iterations and adjustments, whereas any form of physical experimentation will be limited by resources in terms of laboratory availability, materials and time. Further, patients can be protected from avoidable harm in early clinical trials by simulating the interaction between devices or prostheses and the *in vivo* environment. Some of these interactions may be hard to model in the laboratory, as shown by the complexities of mimicking the biomechanics of the hip outlined in this chapter. Serial, small adjustments may be made as and when required with only staff costs to consider.

The method relies on a detailed understanding of the properties of the material under investigation, its geometry and the forces acting on it. The more complex the geometry and less predictable the forces, the harder the model is to generate to an acceptable degree of accuracy. The human musculoskeletal system, with its complex material properties and enmeshed set of interactions between muscles, ligaments and bones at joints presents a significant challenge.

Figure 2.7 shows a FEA model of a hemiarthroplasty model, with a mesh of interconnected points and the von Mises stresses calculated to exist at them.

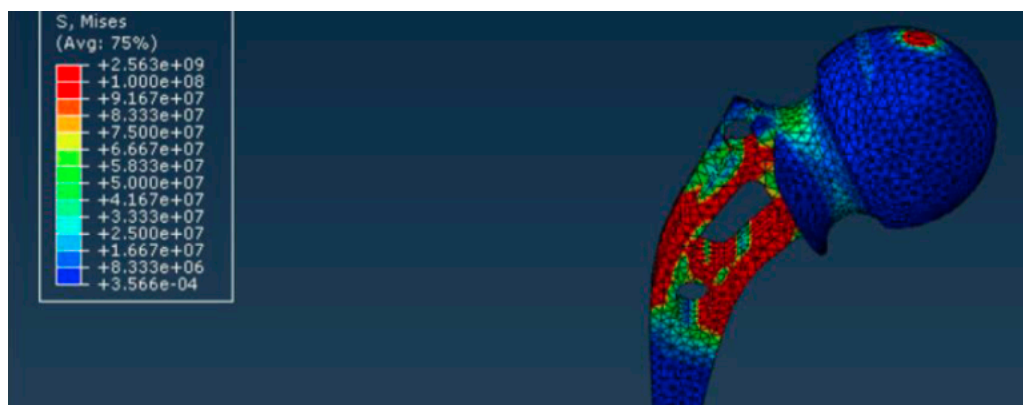


Figure 2.7 – A FEA model of a hemiarthroplasty implant
(Colic et al., 2016)

The technique was first described in the orthopaedic literature in 1972, and early concerns were raised that the initial results of work with the technique did not

reflect the reality of in vivo conditions (Huiskes & Bio, 1983). Prendergast's review paper 14 years later is more optimistic in tone and highlights some reasons why the technique may have become more reliable in the intervening time. Generation of the mesh, the representation of the object by a 3-dimensional series of nodes, could increasingly be based on computerised tomography (CT) scanning (Prendergast, 1997). This increases accuracy of the model whilst reducing the workload, and indeed risk of error from the researcher.

The FEA method has been used in a number of studies of fixation of hip fractures. Helwig et al. used the technique to ascertain the optimal point for the fixation device in the femoral neck and head, for each of four different short IMNs (Helwig et al., 2009). In doing so, they demonstrated the subtle variation between devices in ideal placement of the compression screw, notwithstanding all were devices of the same class, based on the same biomechanical principles. As an aside, this is a very good illustration of the importance of being conversant with specific operative techniques even when familiar with comparable devices.

Seral et al. investigated the stresses in the Gamma and Proximal Femoral nails, identifying higher stresses in the Gamma and postulating that these may be associated with the contemporaneous high rate of periprosthetic fracture and thigh pain described earlier in the chapter (Seral et al., 2004).

Goffin et al. used FEA to model different positions of the compression screw in the femoral head, after Baumgaertners's clinical study of tip-apex distance (Baumgaertner et al., 1995; Goffin et al., 2012). Their findings challenged that of the clinical work, suggesting that inferior screw positions offered most resistance to cut-out, while TAD was not, by contrast, a predictor of successful fixation.

Overall, the method has great potential for implant design and for trying to understand specific clinical problems. It is unlikely it will be able to entirely replace physical biomechanical preclinical studies, but offers great benefit in cost and feasibility for experiments supporting these stages.

2.5 Summary of key biomechanical points in the literature

The stability of extracapsular fractures is based on the degree of comminution and the presence of an adequate buttress, taking into consideration both the posteromedial calcar and the integrity of the lateral femoral wall.

The primary mode of failure of hip fracture fixation is cut-out of the screw or blade from the femoral head. Appropriate positioning of the screw in the femoral head and consideration of the rotational stability of the construct both play their part in avoiding this cut-out.

A number of factors including diameter, curvature, material, slotting and tapering affect the mechanical properties of an intramedullary nail. Failure to consider these properties can result in malreduction, non-union or peri-prosthetic fracture of the nail, with catastrophic failure a concomitant risk.

The forces acting on the hip joint are complex and, *in vivo*, act in tandem with surrounding soft tissue structures. A distinction must be drawn between the joint reaction forces seen in the static hip (a person standing still) and the dynamic forces occurring during motion at the hip joint, such as standing up or sitting down.

Several surrogates for osteoporotic bone exist. These include plastic synthetic bones, which may be a simple cortical shell and cancellous filling or a complex model offering high fidelity when compared with human femora, and acid-treated bovine tibiae.

A number of biomechanical studies have been performed, tending to simulate the largest loads on the hip and hence mainly using axial compression or rotation models. It is generally understood that such experiments add to understanding of the biomechanics of hip fracture fixation, but do not accurately reproduce conditions seen *in vivo*.

Finite element analysis can be used to model fractures and their fixation, with increasing accuracy, but is best considered an adjunct to physical testing.

A new device, the X-Bolt, has been introduced with the aim of reducing failure through cut-out. It has been shown to have increased resistance to pushout and rotation in models not simulating physiological loading.

The WHiTE One study has shown the feasibility of recruiting to a clinical trial comparing the X-Bolt DHS with a conventional SHS, with no cut-outs seen in the X-Bolt group, and the WHiTE Four full study is now in progress.

An X-Bolt nailing system has been introduced, based on the X-Bolt device and a stainless steel intramedullary nail.

Chapter 3: An evaluation of incidence and aetiology of failure of fixation in extracapsular hip fractures

3.1 Introduction

As established in Chapter 1, there has recently been high-quality, RCT research conducted, comparing SHS with IMN constructs in intertrochanteric fractures. This research has tended increasingly towards outcome measures meaningful to patients, recognising the relative rarity of failure of the construct. Catastrophic failure and re-operation rates are generally reported, but these figures do not take into account subtler changes in a patient's day-to-day existence. By using outcome measures driven in the main by mobility and task-based function, one significant aspect left unexplored is that of pain; while many patients with hip fracture regain an acceptable amount of mobility, they do not always do so without pain (Griffin et al., 2015).

The methodology of the rigorous recent trials is largely one where follow-up is within the context of the trial, rather than standard of care (Griffin et al., 2013). This is undoubtedly not least because many such patients are not routinely followed-up from a surgical perspective. Whilst there is no single clear reason for this lack of follow-up, a combination of the burden of work, the high mortality and the expectation that significant problems in a low demand patient would declare themselves probably all contribute to this practice. It therefore makes it hard to generalize the findings of these trials to routine practice in all hospitals managing patients with hip fractures – does a post-operative problem have genuine impact if it only came to light through research follow-up ?

By the same token, the NHFD dataset captures only “bigger” outcomes – a change in mobility denoted by the use of sticks or frames, a change in residential status, re-operation or death. While the data completeness overall is very good for such a complex patient group, the NHFD annual reports have highlighted this long-term functional data is where there is a relative weakness by contrast with the easier to collect in-hospital data (Royal College of Physicians, 2014).

The aims of this study were to:

- Quantify the incidence of post-operative problems in patients with either SHS or IMN fixation of intertrochanteric hip fractures and compare this incidence between subgroups

- Assess technical aspects of the fixation (tip-apex distance and quality of reduction) and patient factors (ASA grade, AMTS, pre-injury mobility) in the context of, age and sex-matched controls who had not re-presented with problems

The null hypotheses were that:

- The incidence of problems does not differ from that reported in the contemporary literature
- No differences in incidence exist between SHS and IMN sub-groups
- No differences exist in technical or patient factors between patients with problems and those without

3.2 Patients and Methods

3.21 Patients

Patients aged 65 years or more undergoing fixation of an intertrochanteric hip fracture at the Royal Surrey County Hospital, Guildford; the Royal United Hospital, Bath or Southmead Hospital, Bristol between 1 January 2013 and 31 December 2015 were eligible for inclusion. The age floor of 65 years relates to the cut-off for inclusion in the National Hip Fracture Database, and hence is an obligate parameter in the context of this study. The inclusion dates were based on a time period during which data volume and integrity in the NHFD was high, having gone from 64 participating units in the first report in 2008 to all 182 eligible units being registered by 2011 (Marsh, 2017; Royal College of Physicians, 2014). The 2014 NHFD report cites some concerns with data completeness, but these related primarily to super-episode information on place of residence at 90 and 120 days post-operatively. The end date to the inclusion period was chosen to leave a sufficient period after surgery for any problem to have become evident and a referral made to a hip surgeon, as data collection commenced in January 2017.

The study was registered as a service evaluation at the three participating sites.

3.22 *Sampling*

A matched case-control methodology was identified as the most appropriate strategy to answer this clinical question. Patients were matched on age, with a 5-year tolerance, and sex, using consecutive patients. A matching ratio of 3:1 of controls to cases was targeted to attempt to control for confounders whilst presenting an acceptable burden of data to be gathered in each hospital. The strengths and weaknesses of this approach are discussed further in this chapter.

3.23 Methods

The patients were identified from each site's National Hip Fracture Database dataset, which permits submitting centres to query their own patient data. In practical terms, this meant filtering the dataset by the fracture type, selecting all those recorded as intertrochanteric. Any patient already recorded on the NHFD as having required re-operation was added to the cohort. For each patient in the dataset, local patient administration systems (PAS) were queried for orthopaedic outpatient episodes occurring after the date of discharge for the fracture. For these episodes, clinic letters were retrieved electronically and the narrative from the consulting clinician checked to ascertain if the appointment related to the hip fracture. Where the appointment was related, the patient became eligible for inclusion in the "Case" cohort and the reason for referral, diagnosis and treatment was recorded. The possible values for reason for referral were coded as: 1, mobility, falls or limp; 2, pain; 3, wound problems; 4, avascular necrosis, osteoarthritis or mal-union; 5, failure of fixation.

For these patients, analysis of AP radiographs at presentation was performed to assign the fracture an AO classification, and of intra-operative fluoroscopy views to determine tip-apex distance. Tip-apex distance was calculated after Baumgaertner et al (Baumgaertner et al., 1995), the sum of the distance between the tip of the screw and the apex of the femoral head on the AP view and the same on the lateral view, having calibrated the image for magnification.. (Baumgaertner et al., 1995), the sum of the distance between the tip of the screw and the apex of the femoral head on the AP view and the same on the lateral view, having calibrated the image for magnification. The apex is defined as the intersection of a line through the centre of and parallel with the femoral neck, and the subchondral bone of the femoral head. Radiograph reviewers were asked also to assess the Garden alignment index (GAI), the angle between the trabeculae of the femoral shaft and that of the neck fragment on AP and lateral radiographs (Garden, 1971). While the GAI was developed to assess the reduction of subcapital fractures, the means of assessing reduction by aligning the trabeculae is common to intra- and extracapsular fractures. A team of one senior and one junior trainee orthopaedic surgeon performed this at each site, with cross-checking and consensus used to ratify findings. A radiograph was deemed as

adequate if the femoral head and neck were fully visible, with margins, in the AP and lateral views. As this study was retrospective, a missing radiograph could only be recorded as missing data. Given the anticipated low incidence of problems, to fail to record a patient with problems could have a significant bearing on the results.

The NHFD was then used to determine the patient's American Society of Anesthesiologists grade (i.e. their level of co-morbidity), pre-injury Abbreviated Mental Test Score (i.e. their cognitive capability) and mobility status. As different centres had been using different mobility ratings within the NHFD dataset, a composite "Best mobility" score was created, which could be derived from any combination of the "Pre-fracture mobility", "Walking ability indoors", "Walking ability outdoors", "Aids to walk indoors" and "Aids to walk outdoors". The best mobility score was rated as: 0, bed- or wheelchair-bound ; 1, mobile indoors but never goes outdoors; 2, goes outdoors with help; 3, mobile outdoors with 2 sticks or a frame; 4, mobile outdoors with a single stick; 5, freely mobile.

Statistical analysis was performed using SPSS version 24 (IBM, Armonk, NY). When comparing the three sites, independent samples Kruskal-Wallis testing was used with the site as the independent variable and age, length of follow-up, best mobility, tip-apex distance, ASA grade and AMTS as dependent variables. This was based on the assumptions of independence of variables between groups (no patient being in more than one group) and a non-normal distribution.

When comparing problem and non-problem groups, Mann-Whitney U testing was used to compare age, length of follow-up and tip-apex distance. Independent samples Kruskal-Wallis testing was used to compare AMTS, ASA grade, best mobility and AO classification. Chi-square testing was used for comparison of operation type and the number of A3 fractures fixed by SHS.

3.3 Results

3.3.1 Patients by site

A total of 4010 patients were featured in the NHFD dataset for the three sites during the study period. Of these, 1260 patients had undergone fixation of extracapsular fractures. A cohort of 57 (4.5%) patients with problems were identified.

The patients eligible for inclusion by site are shown below.

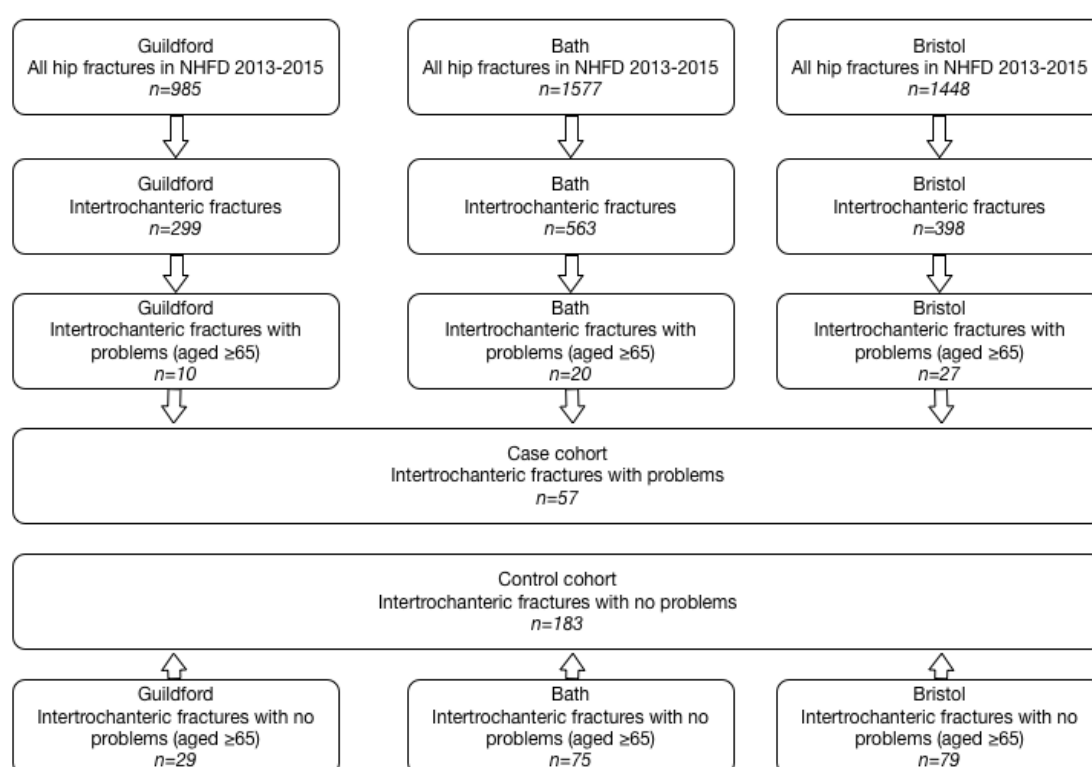


Figure 3.1 – Participant flow diagram

All patients identified had some data available and so remained eligible for inclusion. The rationale supporting this is outlined in the Discussion section of this chapter.

3.32 Patient characteristics

Patient characteristics by cohort are summarized in Table 3.1. The cohorts were appropriately matched for age (79 years in the problem cohort and 81 in the problem-free cohort, $p=0.673$) and sex (82.5% female in the problem cohort and 83.1% female in the problem-free cohort, $p=0.916$). There was a preponderance of patients with low ASA scores (mode ASA grade 2 in the problem cohort *versus* grade 3 in the problem-free cohort) and high AMTS (3 of 57 patients scoring below 6 in the problem cohort *versus* 44 of 183 in the problem-free cohort) , and hence better medical fitness and cognitive function in the case group when compared with control patients who had not complained of any problems ($p<0.0001$ and $p=.0030$), respectively.

Patients had broadly the same pre-injury mobility, with around one third of the patients in each cohort being independently mobile ($p=0.097$), and length of follow-up (25.56 days, SD 11.56 days, *versus* 22.34 days, SD 14.39 days, $p=0.143$).

Table 3.1 – Patient characteristics by cohort

		Case	Control	p value
Total		57	183	
Gender	Male	10	31	.916†
	Female	47	152	
Age		81.21 (65 – 98, SD 8.174)	81.62 (65 – 99, SD 71.21)	.673††
ASA grade	1	4	3	<0.0001†††
	2	36	45	
	3	16	104	
	4	1	30	
	5	0	1	
AMTS	0	0	17	.003†††
	1	0	6	
	2	0	6	
	3	1	4	
	4	2	4	
	5	0	7	
	6	2	7	
	7	2	6	
	8	4	6	
	9	10	19	
	10	36	22	
Best mobility	Bed- or wheelchair-bound	2	7	.097†††
	Never goes outdoors	1	6	
	Mobile outside with help	2	22	
	Mobile with 2 sticks or frame	5	18	
	Mobile with single stick	8	25	
	Independently mobile	21	55	
Follow-up (months)		25.56 (0 – 47, SD 11.56)	22.34 (0 – 50, SD 14.39)	.143††

3.33 Surgical characteristics

Surgical characteristics are shown in Table 3.2.

A similar distribution of AO sub-type was seen between both groups ($p=0.146$). This information was unavailable in 16 case and 18 control patients.

The researchers at all sites reported that it was impossible to measure the Garden Alignment Index in the majority of patients, as the only imaging available was from intra-operative fluoroscopy which lacked the resolution to adequately demonstrate the bony trabeculae of the proximal femur.

The distribution of operations was the same across both groups with 8 of 57 (14%) of patients undergoing intramedullary nailing in the problem cohort and 22 of 183 (12%) in the problem-free cohort ($p=.501$). The case group had a significantly higher proportion of SHS fixation for A3 fractures (7 of 11) than the control group (3 of 13) ($p=0.045$) (Figure 3.2).

The tip-apex distance was significantly lower in the problem-free cohort, at 15.32 mm versus 17.21 mm ($p=0.005$), but both these values lie well within Baumgaertner's advocated range.

The most frequently occurring problem was failure of fixation (22 of 57) with mobility problems and avascular necrosis, osteoarthritis or malunion equally common in 11 of 57 patients.

Table 3.2 – Surgical characteristics by cohort

		Case	Control	p value
AO	31A1	13	45	.146 ^{†††}
Classification	31A2	27	107	
	31A3	11	13	
	missing	16	18	
TAD (mm)		17.21 (3 – 28, SD 5.378)	15.32 (5 – 37, SD 5.047)	.005 ^{††}
Operations	SHS	49	156	.501 [†]
	IMN	8	22	
	missing	0	5	
A3 fractures fixed with SHS		7 of 11	3 of 13	.045 [†]
Problems	Mobility, falls or limp	11		
	Pain	8		
	Wound problems	5		
	AVN, OA or mal-union	11		
	Failure of fixation	22		

(† chi-square test, †† Mann-Whitney U test, ††† Independent samples Kruskal-Wallis test)

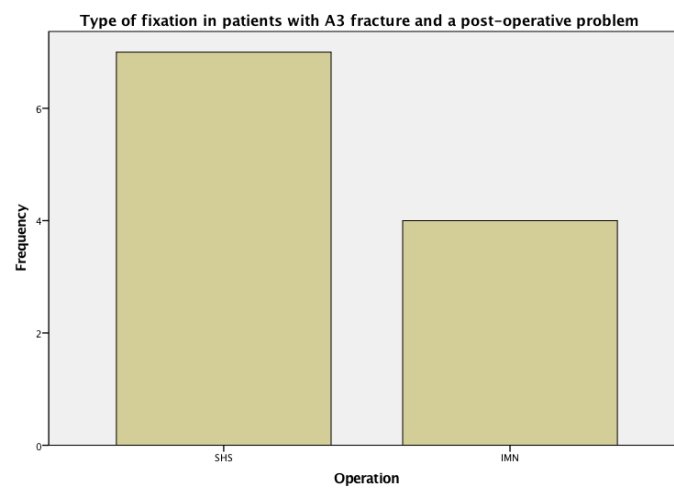
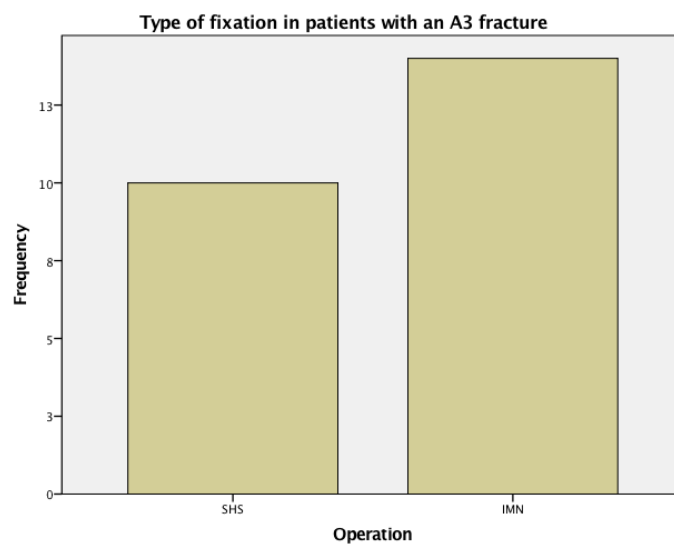
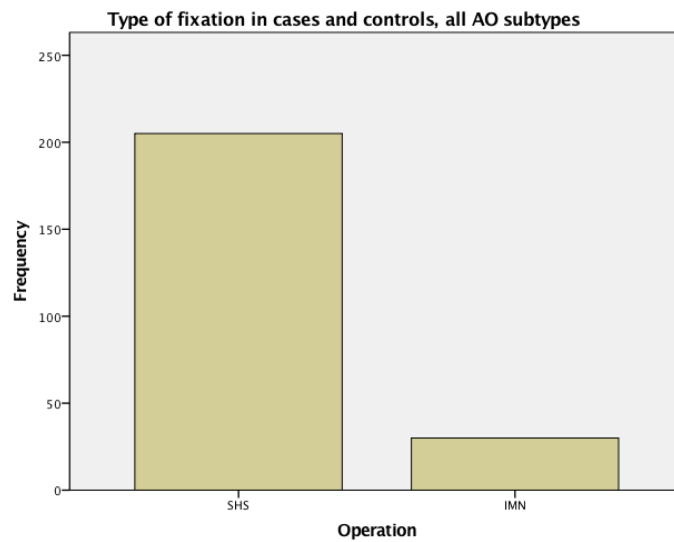


Figure 3.2 – Type of fixation in (a) all cases and controls (b) all cases and controls with A3 fractures and (c) cases with A3 fractures

3.4 Discussion

The incidence of problems was low, in the region of 5% for all problems and 2% for failure of fixation. This is in keeping with previous work by this team and other contemporaneously published rates for the UK (Page et al., 2016). (Page et al., 2016). This study builds on that work by exploring more patient characteristics and focusing less on temporal trends in usage of different fixation devices.

The higher tip-apex distance in the problem group resonate with concepts existing in the literature (Baumgaertner et al., 1995) but it is hard to be confident that these do not represent a type 1 error in the context of a relatively small sample size and values under the widely accepted threshold of 25 mm. Many surgeons would not, however, accept a fixation only just within the standard 25 mm TAD and hence this finding may suggest surgeons should aim for as close to zero as possible, rather than just any sub-25 mm distance.

The distribution of ASA grade and AMTS may reflect an increased likelihood of patients with less comorbidity and less cognitive impairment to report problems such as pain and reduced mobility. In patients whose health and ability to communicate is worse, it may be that a problem needs to be greater to become of clinical significance or needs to be overtly evident to a carer if a patient cannot communicate it. On this basis, a radical change in mobility or an infected wound is likely to be detected, whereas hip pain which causes distress without causing a gait disturbance, or a loss of confidence in gait may only be unmasked in patients cognitively capable of reporting it.

The high incidence of problems in A3 fractures fixed by SHS is an important one. Whilst this study is not adequately powered for more complex inferential statistics such as multivariate analysis, this trend is noteworthy both for its position in relation to current guidance, which suggests a preference for IMNs in these fractures (National Clinical Guideline Centre, 2017), and the fact that poor outcomes are evident in this subgroup. A multivariate model addressing all the many relevant independent variables would, in the context of small effect sizes across all fracture subtypes, likely require several thousand patients. More crucially, this complex model would not directly address the question at hand,

namely whether there is a population of patients who experience problems after fixation of their extracapsular hip fracture.

This study demonstrates that there still exists a sub-group of trochanteric fractures associated with poorer outcomes if fixed with an SHS, and hence provides some further support to existing literature suggesting that there is a valid place for the intramedullary fixation of unstable trochanteric fractures (Socci et al., 2017; Jacob et al., 2017). This work also demonstrates that problems can still occur after fixation with an IMN, and so a new device may still be able to confer a benefit. The absence of AO classification in 34 fractures and pre-fracture mobility data in 69 patients limits the impact of the findings relating to those factors. Nevertheless, the primary goal of this study was to quantify the incidence of post-operative problems and it achieves this.

The inability to derive the Garden Alignment Index from fluoroscopy is a significant negative – it removes any assessment of the adequacy of reduction from this study, which is regrettable, but articulates a clear learning point for any future researchers planning to work in this area. Any research methodology for this will have to encompass either additional formal radiographs, usually only the standard of care in some centres for intramedullary nailing, or make use of a technique such as radiostereometric analysis (RSA) to gain information on the reduction of a fracture. As the reduction of an extracapsular fracture is so paramount in successful fixation, any further studies should consider this a key information requirement. It may also be that a better means of assessing reduction exists, as the GAI was described as a measurement by which the reduction of intracapsular fractures was assessed. There is also an argument to support not measuring the angles relating to reduction, but rather to have surgeons state whether a reduction is adequate or not. While this may intuitively seem very subjective, it is reflective of exactly how decisions are made in clinical practice where the fixation stage of the operation only proceeds once the surgeon is convinced they have attained the optimum reduction.

The decision to use matched cohorts was intended to reduce the heterogeneity of the samples, by ensuring age and sex were broadly similarly distributed between groups. The decision was taken early on to avoid trying to match any other variables as the pool of potential participants would probably not support

sufficient combinations to match specific combinations of 3 or more variables. It is unlikely, however, that an unmatched cohort would have had a significantly different median age or different female to male patient ratio and it may be that this matching strategy actually reduced the power of the study. Failure to detect an effect of exposure has been described when using a matching method, notably by authors such as Marsh et al. who described how this method led to failure to detect a significant increase in the incidence of leukaemia in workers at a nuclear fuel processing plant (J. L. Marsh, 2002). Other authors have applied matched and unmatched methodologies to the same dataset and found broadly similar results, but with diminished statistical power in the matched study (T. Faresjö & Å. Faresjö, 2010). Future work in this area may be better served by simply describing the populations encountered, rather than matching them.

The assumption that patients experiencing problems with their fixation would return to their treating hospital is also one open to question. A more robust methodology would have been to access population-level data; Nedza et al. were able to report Emergency Department attendances in arthroplasty patients, including those undergoing total hip replacement for fracture, by using a state-wide billing system to capture re-presentation anywhere in the state (Nedza et al., 2017). This type of methodology relies on high-quality centralised data, which in the UK may be obtained through Hospital Episode Statistics (HES) information. Such investigation was, however, beyond the resource constraints of this project. The argument that patients will seek help from or be referred back to their treating centre is one which must be counter-balanced with the possibility of patients explicitly seeking a second or different opinion from a different centre due to discontent with outcomes. This remains a perennial problem in retrospective research and is well-described in the literature, without a robust solution apparent (Morris et al., 2011; Zmistowski et al., 2013).

Further discussion of these results in the context of the wider project and literature can be found in Chapter 6. The next chapter addresses the stiffness of stainless steel nails by comparison with those manufactured from titanium.

Chapter 4: A comparison of 3-point bending in stainless steel and titanium alloy nails

4.1 Aims and objectives

As outlined in Chapter 2, the mechanical properties of intramedullary nails vary according to a number of factors. Diameter, thickness, taper, material and length have all been shown to affect biomechanics and risk of periprosthetic fracture in the proximal femur (Eveleigh, 1995; Russell, 2011; Henley et al., 1993). A particular area of concern has been the use of stainless steel in these devices, which was common in the early generations of modern cephalomedullary nails with which a high incidence of periprosthetic fractures were seen. This perceived increased risk of fracture prompted adoption of titanium alloys for the manufacture of these nails, with its relatively lower Young's modulus and hence increased elasticity, closer to that of bone. Often unacknowledged, however, when comparing these early devices (an example of which is shown in Figure 4.1) with contemporary counterparts, is the fact that a number of other design features changed and these may contribute to the modern, substantially lower rate of complications such as periprosthetic fracture. Such changes include a reduction in diaphyseal diameter, a long taper of the implant at its distal end, fluting of the device and moving the distal locking screw to a more proximal location on the nail to avoid an abrupt stress riser at the end of the device. This re-design process is not, however, evident in the literature and may well be documented only in technical notes and commercially protected white papers.

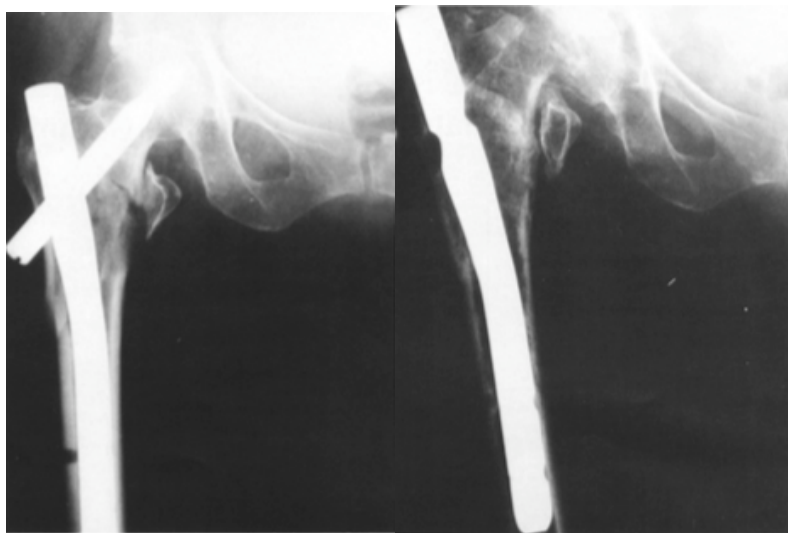


Figure 4.1 – Gamma nail with (a) failed distal locking and (b) subcapital fracture after removal of compression screw

(Williams & B. C. Parker, 1992)

On this basis, therefore, the anecdotal perception that the failures of early nails proved the unsuitability of stainless steel for their manufacture is at the very least open to question.

The aim of this experiment was to compare the stiffness of femoral models instrumented with stainless steel or titanium nails, using a 3-point bending method. This was to demonstrate to surgeons unwilling to consider a clinical trial of a steel IMN on the basis of its stiffness that it need not necessarily differ from that of a titanium IMN which they would be willing to implant.

The null hypothesis was that there would be no significant difference in stiffness between nails manufactured from stainless steel or titanium.

4.2 Materials and models

4.2.1 *Materials*

Twelve composite synthetic right femoral models were used for all experiments (Orthobone Premium, 3B Scientific GmbH, Germany). These models are 470 mm long, have a medullary diameter of 8mm and a femoral head diameter of 43mm. They feature cortical and cancellous analogues, and weigh 200g.

The selected nails were the stainless steel X-Bolt nail, and a modern standard of care titanium device, the Zimmer Biomet Affixus.

The X-Bolt nail system (X-Bolt Orthopaedics, Dublin, Ireland) comprises a stainless steel, 14.5mm proximal and 10mm distal diameter IMN, with a 4° valgus angulation, 130° femoral neck angle and 2m bend radius. It is locked proximally by means of the novel X-Bolt expanding bolt device and distally by 5mm diameter screws.

The Affixus nail system (Zimmer Biomet, Warsaw, IN) features titanium nails with a 15.6mm proximal diameter and 9mm, 11mm or 13mm distal diameters. It also has a 4° valgus and offers 125° or 130° femoral neck angles. It has a 1.8m bend radius. It is locked by means of 5mm diameter screws. The 11 mm nail was available for testing in this experiment.

Figure 4.2 shows schematic drawings of both the Affixus and X-Bolt nails.



Figure 4.2 – Schematic drawings of (a) Affixus and (b) X-Bolt nails

4.22 *Construction of the synthetic bone jig*

Micro-computed tomography (Nikon XT 225 ST, Nikon Metrology, Hertfordshire, UK) was used to obtain three-dimensional images of the synthetic femur, from which a software model was created (ScanIP, Simpleware Ltd, Exeter, UK). This was used in turn to design components to restrain the synthetic bone on the jig, and to guide the cuts required to create the necessary fracture line. The custom components were then produced using additive manufacturing, or 3D printing (UpBox, Tiertime, Beijing), from acrylonitrile butadiene styrene (ABS) before being secured to an aluminium plate (Figure 4.3).

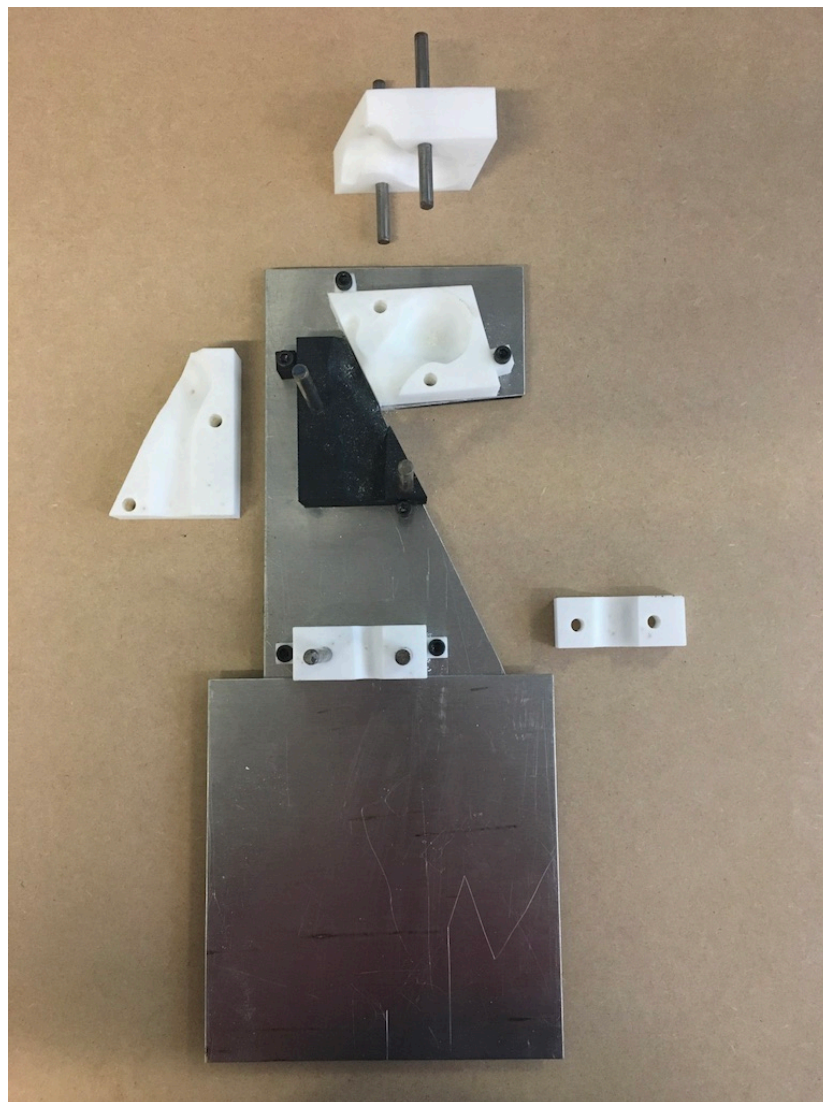


Figure 4.3 – The 31A2 osteotomy jig

Baseplate with custom restraints for synthetic femur and an oblique cut-out on the right-hand side to guide the saw blade in production of a 31A2 fracture pattern. The components not attached are those which restrained the Orthobone from above, and secure into the matching bottom components to which they are adjacent.

4.23 *Creation of the uniform fracture model*

The posterior condyles of the synthetic femur were resected without the use of a cutting guide, to permit clearance of the distal femur from the saw table when attached to the jig (Figure 4.4).



Figure 4.4 – Distal femoral resection in (a) Postero-anterior and (b) lateral plane

The femoral model was then restrained in the jig, and a bandsaw used to make an intertrochanteric cut, and a second cut used to form a wedge osteotomy at the medial calcar (Figure 4.5).

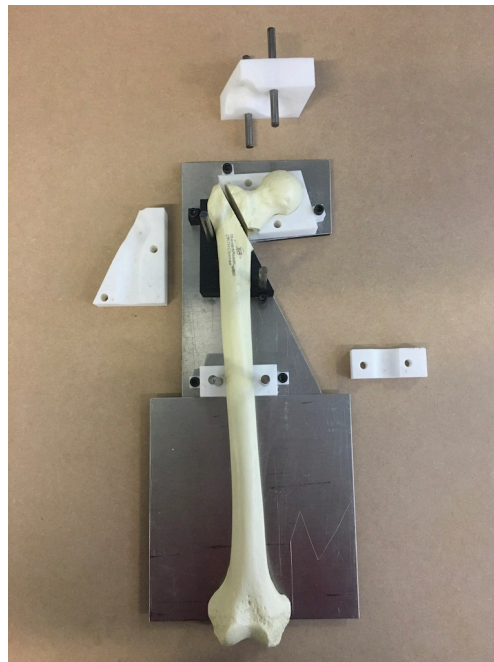


Figure 4.5 – Sawbone in jig post-osteotomy

The resulting model was analogous to a 31A2 hip fracture, with the defect from the excised fragment conferring the inherent instability implied by this fractured type (Figure 4.6).

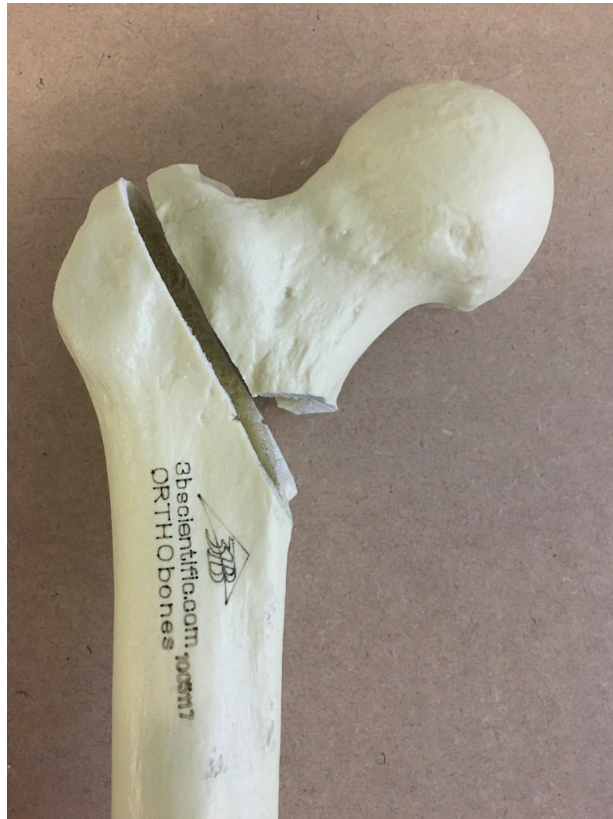


Figure 4.6 – Sawbone post-osteotomy, with intermediate lesser trochanter fragment discarded

4.24 *Creation of the instrumented femur model*

The fractured femur model was instrumented with the IMNs. The fracture was reduced and held in a vice, analogous to the reduction performed on a traction table. An entry point was then established just medial to the trochanteric tip; in practical terms, this was facilitated by a predictable defect in the plastic, likely used for the extrusion of material during manufacturing of the model. This entry point was opened by means of the entry reaming awl, giving sufficient capacity in the metaphysis to accommodate the larger (between 14 mm and 16 mm) proximal diameter of the IMNs.

Flexible reamers were then used to ream the medullary canal to 13 mm diameter. The nail was then introduced into the femur, using the marking of the centre of the femoral neck as a guide against which the proximal locking hole could be aligned. The proximal locking device was then introduced using a guide wire, cannulated drill and then the proprietary instrumentation required. In the case of the X-Bolt, this was a bone-crushing device and then a torque-limited screwdriver to deploy the cruciform arms of the X-Bolt.

Distal locking was performed using a machine press to drill the bi-cortical hole, through the nail, and standard M5 machine bolts with Nyloc nuts used in place of proprietary screws (Figure 4.7).



Figure 4.7 – X-Bolt nail locked (a) proximally and (b and c) distally

The X-Bolt model was created with 10 mm distal diameter nails, the universal size in which they are produced, while the Affixus model used 11 mm implants, those available from the manufacturer for the purposes of these experiments.

The intention was to create eight models in each group. Eight models were successfully constructed using the X-Bolt nail, but only four with the Affixus system. The Affixus models used the same nail for each model, one after another, whereas the X-Bolt system had a number of nails available and so those models were made as a batch. The causes of failure of instrumentation were cortical breach from the flexible reaming system, and periprosthetic fracture occurring during nail insertion (Figure 4.8).



Figure 4.8 – Orthobone model showing common mode of failure, a periprosthetic fracture in the proximal third of the femur

4.3 Methods

4.3.1 Testing

Testing was performed using an Instron ® 5967 electromechanical universal testing machine (Instron, High Wycombe, UK) fitted with a 30 kN load cell. The specimen was centred on a jig with points of contact 102mm apart, the distance set by the fixed nature of the longest jig on which the femoral model would rest without displacing due to its curvature. The femur was oriented such that the model rested naturally on the flattened aspect formed by the edge of the linea aspera, and it was not further secured (see Figure 4.9 below).

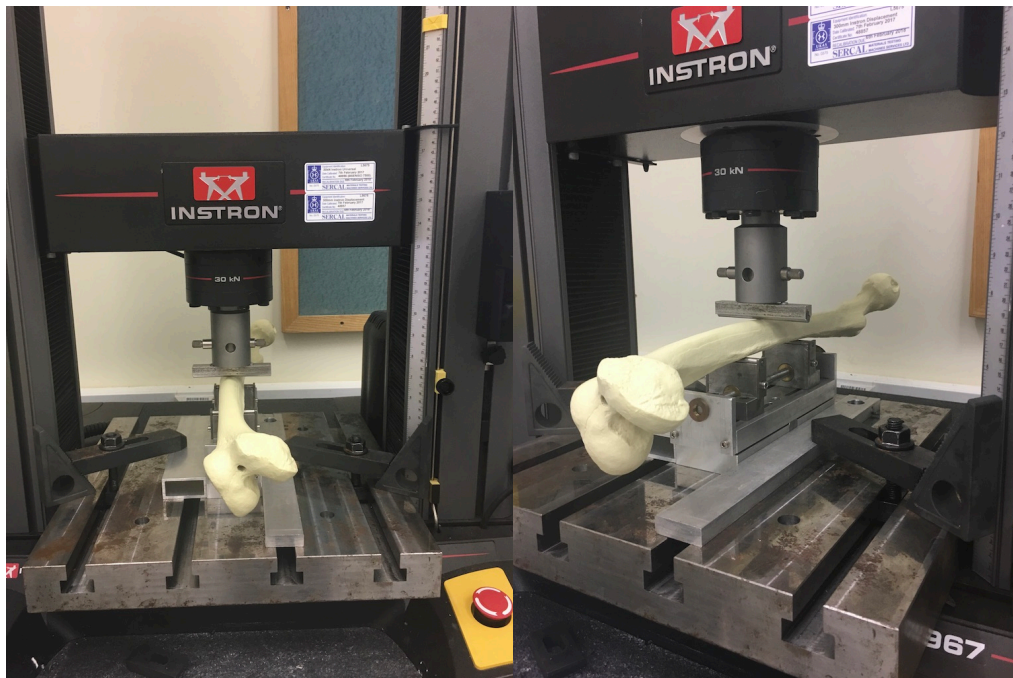


Figure 4.9 – Femoral model mounted for testing of 3-point bending

The proprietary Bluehill ® (Instron, High Wycombe, UK) software designed for the test system was used to create a test profile to apply 500 N of load progressively with a maximum rate of extension of 0.8mm/minute, sampling extension of the composite specimen at a rate of 100Hz. This load was chosen to mimic the calculated loads occurring in the mid-femur during activities of daily life (D'Angeli et al., 2013). A slow strain rate and high sampling rate were chosen as to ideally avoid catastrophic failure of the previously un-tested model, and also to ensure that maximum data was collected before such a failure should it occur.

Once a load of 500 N was achieved, the load cell automatically withdrew until no load was applied.

This test was performed first with an intact, un-instrumented model femur, and then with the models in each group. The uninstrumented femur was used to give context to the results seen in the two instrumented models.

4.32 Calculations and statistical methods

The stiffness of the construct was calculated using the change in displacement between two points in the load cycle. These points, 250 N and 450 N, were selected to represent the part of the loading cycle where any initial settling of the specimen will have taken place, and hence any displacement should be due solely to the material properties of the model. This can be seen in the chart below; the line between these points is relatively straight, representing a constant rate of change.

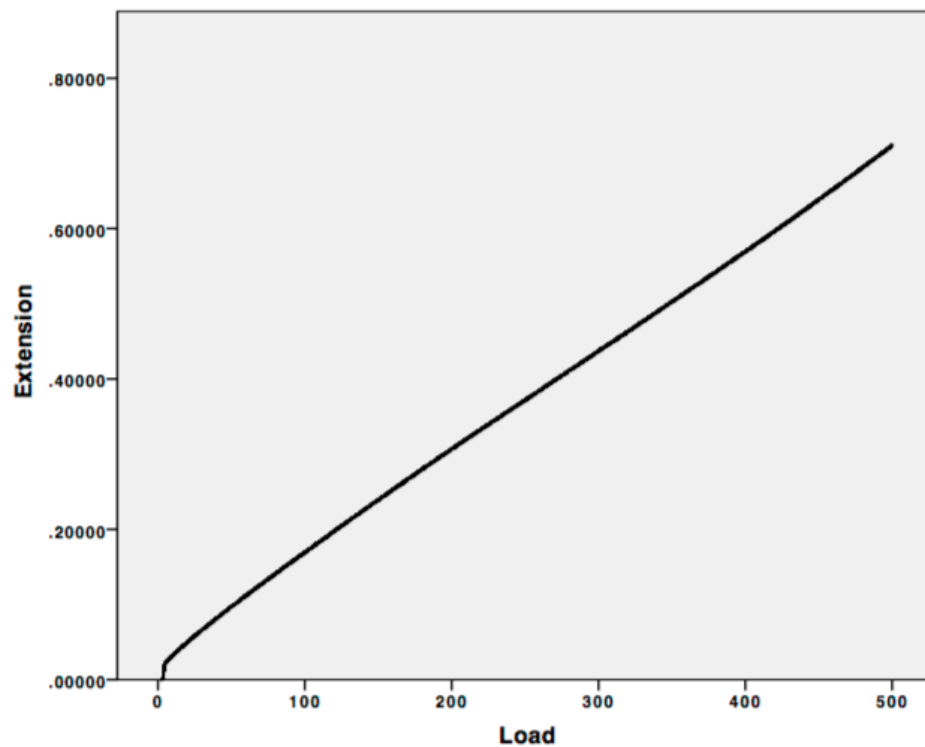


Figure 4.10 – Loading cycle for stiffness testing of femoral models (Extension in mm, load in N)

Stiffness was calculated using the equation

$$\text{Stiffness} = \frac{\text{Change in load}}{\text{Change in extension}}$$

Experimental data were tested for normality by the Shapiro-Wilk test. As this showed them not to be normally distributed, the non-parametric Independent Samples Mann-Whitney U test was selected for comparison of stiffness between experimental groups, and the Independent Samples Kruskal-Wallis test for

comparison of stiffness between groups including the uninstrumented femur. All statistical testing was performed with SPSS version 24 (IBM, Armonk, NY).

4.4 Results

The summary of stiffness data gathered during this experiment is shown in Table 4.1. Table 4.2 shows that the data was not normally distributed, upon which basis the remainder of the statistical tests were selected.

There was no significant difference in stiffness between the 10 mm stainless steel X-Bolt nail and the 11 mm titanium Affixus nail ($p=1.000$), as shown in Figure 4.11 and Table 4.3.

When compared with an uninstrumented, intact synthetic femur, there was also no significant difference in stiffness ($p=0.564$), illustrated in Figure 4.12 and Table 4.4.

Table 4.1 – Change in extension between 250N and 450N of load, with calculated stiffness values

Group	Model	Time (s)	Extension (mm)	Load (N)	Time (s)	Extension (mm)	Load (N)	Change in load (N)	Change in extension (mm)	Stiffness (N/mm)
X-Bolt	1	279.01	0.37	250.00	478.81	0.64	450.00	200.00	0.27	750.55
X-Bolt	2	373.01	0.50	250.02	740.21	0.99	450.00	199.98	0.49	408.45
X-Bolt	3	286.62	0.38	250.01	484.27	0.65	450.02	200.02	0.26	758.93
X-Bolt	4	252.13	0.34	250.01	431.09	0.57	450.03	200.02	0.24	838.22
X-Bolt	5	275.11	0.37	250.00	470.98	0.63	450.01	200.01	0.26	765.99
X-Bolt	6	265.62	0.67	250.01	465.33	0.94	450.04	200.03	0.27	751.11
X-Bolt	7	263.77	0.35	250.05	453.41	0.60	450.01	199.96	0.25	790.90
X-Bolt	8	358.86	0.48	250.01	579.69	0.77	450.02	200.02	0.29	679.18
Affixus	9	412.21	0.55	250.00	610.89	0.81	450.00	200.00	0.27	754.45
Affixus	10	404.66	0.54	250.01	598.18	0.80	450.01	200.00	0.26	775.51
Affixus	11	395.43	0.53	250.01	586.89	0.78	450.01	199.99	0.26	783.59
Affixus	12	406.21	0.54	250.01	654.31	0.87	450.00	199.99	0.33	604.57
Uninstrumented	13	323.59	0.43	250.00	514.32	0.69	450.03	200.03	0.25	786.51

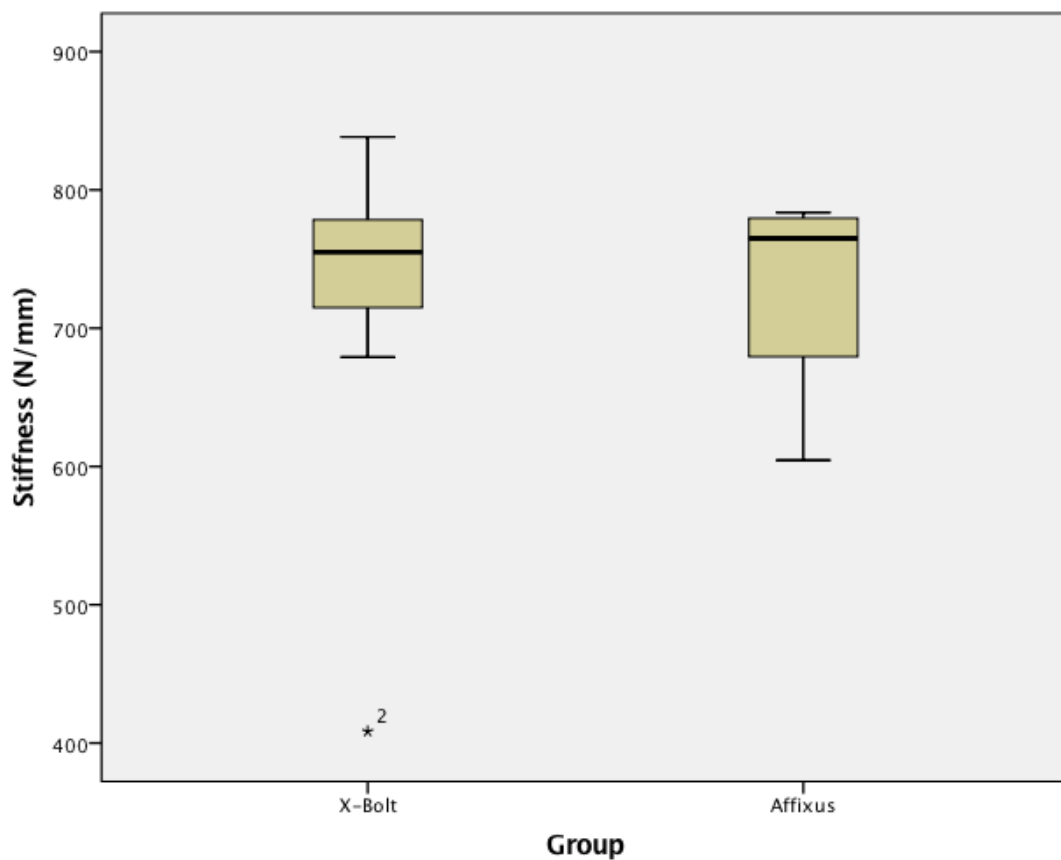


Figure 4.11 – Stiffness of model by group, without un-instrumented femur

Table 4.2 – Shapiro-Wilk tests for normality of distribution of stiffness in X-Bolt and Affixus models

Group	Statistic	df	p value
X-Bolt	.723	8	0.004
Affixus	.753	4	0.041

Table 4.3 – Independent Samples Mann-Whitney U Test for stiffness of X-Bolt and Affixus models

Group	Mean stiffness (N/mm)	SD	p value
X-Bolt	717.92	132.68	1.000
Affixus	729.53	84.20	

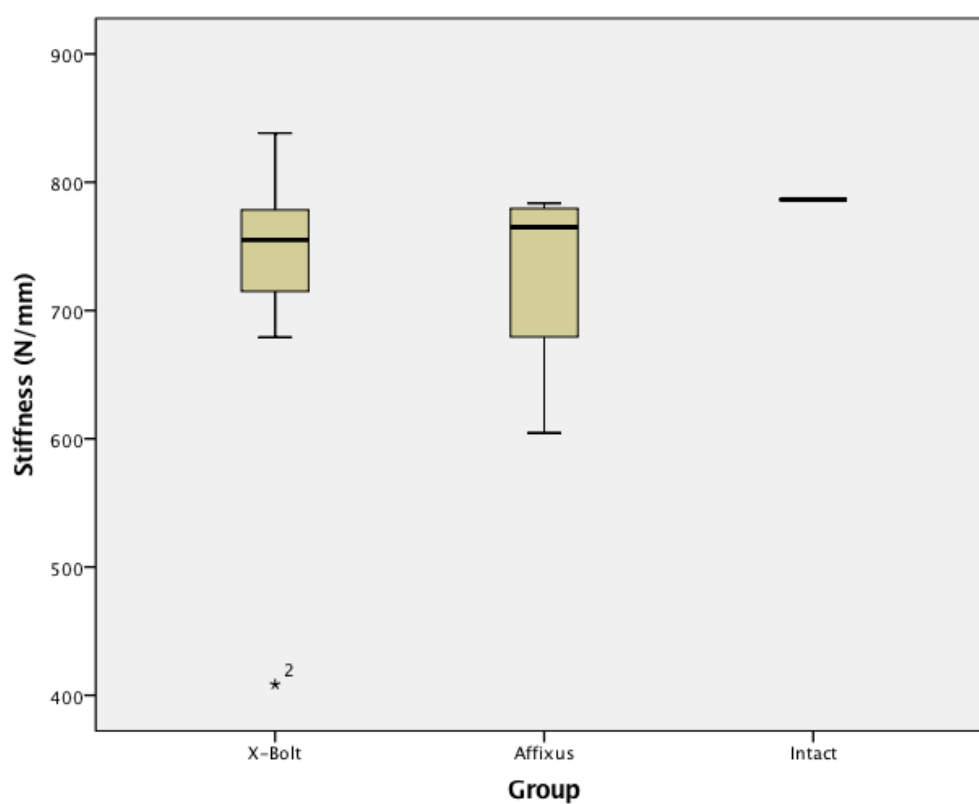


Figure 4.12 – Stiffness of model by group, with single un-instrumented femur for comparison

Table 4.4 – Independent Samples Kruskal-Wallis Test for stiffness of X-Bolt and Affixus models, with single un-instrumented femur for comparison

Group	Mean stiffness (N/mm)	SD	p value
X-Bolt	717.92	132.68	0.564
Affixus	729.53	84.20	
Intact femur	786.51*		

* the stiffness value for the intact femur is expressed as the single calculated value, rather than the mean

4.5 Discussion

The results of this 3-point bending experiment suggest that stainless steel is not inherently unsafe to use as a material for the manufacture of intramedullary devices when appropriate regard is paid to mitigating the differences in metallurgical and material properties between it and current titanium alloys. The stiffness recorded in the experiment is similar to that reported by Wang et al. in their biomechanical characterisation of a new femoral intramedullary nail (G. Wang et al., 2008). There is little other comparative data to be found in the literature, as many other studies tend to focus on axial or torsional loading. Earlier work such as Kyle's characterisation of a novel interlocking femoral nail (Kyle et al., 1991) evaluated such a different form of nail that its results are not readily translatable. It is also noteworthy that this experiment, similar to Cristofolini's characterisation of composite femora (Cristofolini et al., 1996) used 4-point bending rather than 3-point, citing American Society for the Testing of Materials guidance. Any further such experiments may be better performed via this method for more ready interpretation in context of other work.

The similarity in stiffness between both instrumented models and the intact femur reflects that both nails, in different ways, achieve good restoration of the biomechanical properties of the femur. This should help patients achieve their goal of early regain of weight-bearing status and functional mobility. Such restoration of native biomechanics is a key concept in orthopaedic trauma surgery in general, and early remobilisation is the cornerstone of treating fractures in the elderly (Kammerlander et al., 2018).

There exists a difference in design philosophies between the two nails, in that the X-Bolt nail, available only with a 10 mm distal diameter, is less focused on achieving a large proportion of its surface area being in contact with the cortex. The Affixus system, however, has more options of distal diameter in the range and the operative technique is predicated on the surgeon achieving the maximum safe diameter. The similar stiffness shows, however, that the end result is broadly the same in terms of rigidity of fixation. Eveleigh's review of biomechanical experiments characterising intramedullary nails (Eveleigh, 1995) noted that the choice between solid or tubular nails was the main factor significantly altering stiffness, with the concomitant substantial effect on second moment area.

The synthetic femur model is relatively difficult to create, in terms of successful implantation of the intramedullary nails. A number of broken models meant that the comparison was performed between 8 X-Bolt and 4 Affixus nail models. This unbalanced group size is sub-optimal, but a constraint of the funding of the project. Any future project should consider different synthetic femora with more capacious canals, or smaller diameter nails to mitigate the risk of fracture. A technique modification could and ideally should also have been made by passing the long guide wire out through the distal end of the synthetic femur, offering reassurance that the implant would then be passed centrally thereby reducing the risk of fracture. While the synthetic bones, with their lack of viscoelastic properties, presented a challenge for this experiment, they do have the benefit of standardisation, giving both confidence that results should be reproducible and that processes for manufacture and testing can be standardised. The alternative of cadaveric bone, whilst seeming to have greater face validity, is in fact less straightforward. Subtle anatomical differences, as well as almost certain variability in bone mineral density, would make these far more heterogeneous models. It is likely that CT analysis of each specimen would be required to quantify these, significantly raising the complexity of the project. The use of block synthetic material without the structure of synthetic bone would not have realistically modelled the implant-bone interaction as it would have lacked either the stiffness of cortical bone or the friction imparted by cancellous bone. By the same token, simply testing the bending of the implants themselves would be to remove all context from the experiment, undoubtedly affecting its face validity.

The missing data from the broken specimens should likely be considered to represent a surgical failure, in that a device could not and would not be implanted and left in a patient in the way that these models took shape, and so while losing the data to analysis is frustrating, the results would not be generalisable to an *in vivo* situation.

The design of this modern stainless steel nail is fundamentally different to that of its early generation counterparts and the use of this metal should not evoke a reflex aversion to considering implantation of this nail. Once more, a more detailed discussion can be found in Chapter 6. The next chapter deals with the resistance to torsion of the novel expanding bolt itself.

**Chapter 5: A comparison of resistance to torsion
between the X-Bolt device and a conventional
compression screw**

5.1 Aims and objectives

As outlined in Chapter 2, the hip is subject to a number of static and dynamic forces. The X-Bolt has already been tested in axial compression, the force primarily responsible for inducing varus cut-out, and in resistance to rotational displacement (Gosiewski et al., 2017; Gibson et al., 2012)..

Bergman et al. (Bergmann et al., 2016) demonstrated that large forces act in the longitudinal axis in relation to the long axis of the femur, i.e. anterior and posteriorly directed forces acting on the femoral head which in an unrestrained model would result in rotation of the nail within the femur. Given it is distally and proximally locked, however, these forces may also move bone around fixed components, i.e. cause loss of reduction of the fracture. This has the potential in osteoporotic bone to cause cavitation of the head over a number of load cycles. This cavitation risks in turn that the compound movements involved in standing or sitting, i.e. movement into or out of hip flexion induce rotational forces on the femoral head with respect to the femoral neck. As discussed earlier, these rotational forces carry a risk of failure of fixation and hence any design minimizing them may confer a clinical benefit (Lustenberger et al., 1995; Lenich et al., 2011).

The aim of the experiment described in this chapter was to compare resistance to torsion, by measuring the displacement induced at the end of each cycle of loading.

5.2 Materials and models

The 12 models already created for the 3-point bending testing and described in Chapter 4 were used for this experiment.

The X-Bolt device is manufactured from stainless steel and is described fully in Chapter 3 For this experiment, the 90 mm90mm version was used, in keeping with the length of the femoral neck in the plastic bone models.

The Affixus system uses a single or double compression screw, made from titanium alloy. In keeping with the manufacturers' guidance, the single screw was used in the current experiment as the 31A2.1 fracture is relatively stable. A 90 mm90mm screw was used.

5.3 Methods

The instrumented model was restrained in the jig used to make the cuts, secured on the femoral side of the fracture by multiple clamps. The femoral head side of the fracture was unsecured and hence permitted to move freely on application of load.

A 60 mm outer diameter Trident ® PSL acetabular component (Stryker, Kalamazoo, MI) was secured to the Instron ® electromechanical universal testing machine (Instron, High Wycombe, UK) by passing an M5 stainless steel bolt through the central hole used for attachment of the component to the introducer handle during surgical implantation and impaction and clamping this in place with 5x M5 stainless steel nuts. The orientation of the nuts was aligned, giving a square edge by which to secure the component. The final setup is shown in Figure 5.1.

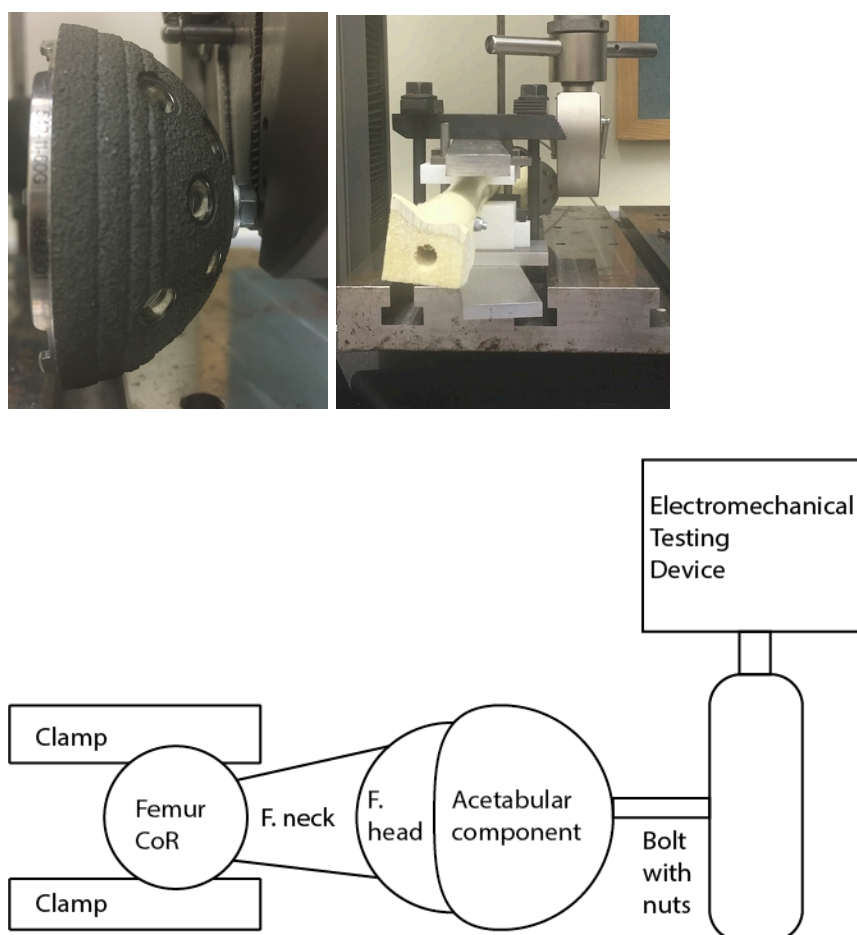


Figure 5.1 – Electromechanical testing setup showing (a) acetabular component mounted in jaws (b) specimen engaged in tester and (c) schematic of testing apparatus.

CoR – centre of rotation; F. neck – femoral neck; F. head – femoral head

The testing cycle was created based on the loads during different activities, demonstrated by Bergmann's work (Bergmann et al., 2016). In these, torsion moments around the longitudinal axis of the femur of 0.93 Nm were seen at the maximum values in stance, and -0.57 Nm in sitting down. With a moment arm of 48 mm existing between the centre of the femoral nail and the centre of the femoral head, these resulted in calculated forces of 19.375 N and -11.875 N respectively. Given this work was performed in vivo and hence these loads were shared between the other static and dynamic stabilizers of the hip joint, the decision was taken to test to a factor of 10 of these values for a worst-case scenario, giving values of 193.75 N and -118.75 N. This is in keeping with another Bergmann paper, which found peak hip joint contact forces in excess of 870% of normal when one of the participants stumbled, reflecting an unexpectedly high peak load (Bergmann et al., 1993). The orientation of the axes were matched between the Bergmann study and the setup of the electromechanical testing device. Ten cycles of a posteriorly-directed force of -193.75 N followed by an anteriorly-directed one of 118.75 N were applied, with displacement measured throughout the cycle. This cycle is shown in Figure 5.2.

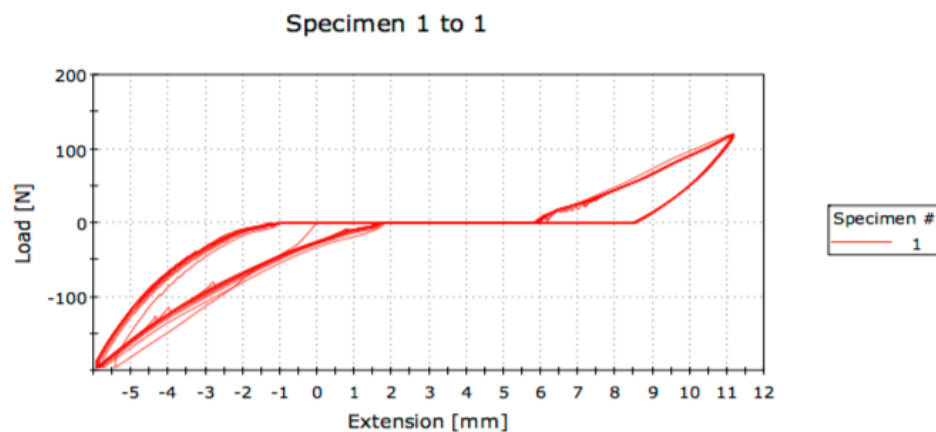


Figure 5.2 – Loading cycle for torsional testing of femur models

Given the difference in size between the model's femoral head and the inner diameter of the acetabular component, a gap existed in the test apparatus. This therefore meant there were phases of the cycle where effectively no load was being applied, but extension was being measured, with the loading phase only occurring once the femoral head contacted the acetabular component. To

calculate peak displacement, the extension at the point at which the load started to increase (because the femoral head had come into contact with the acetabular component) was subtracted from the displacement at peak load.

Data were tested for normality using the Shapiro-Wilk test and based on the result of this, Independent Samples Mann-Whitney U testing used to compare displacement between implants and Independent Samples Kruskal-Wallis testing used to compare displacement between implants in the context of the intact, uninstrumented femur. All statistical testing was performed using SPSS version 24 (IBM, Armonk, NY).

5.4 Results

Specimen 12, the final Affixus specimen, became unstable during the 8th cycle of testing, at which point the test was discontinued due to the safety risk of catastrophic failure under load.

Table 5.1 shows the mean displacement in stance and sitting tests for all specimens. The full dataset of peak values and calculated displacement can be found in Appendix 2.

Table 5.2 shows the results of Shapiro-Wilk testing of normality on this data.

Table 5.3 shows mean displacement in stance and sitting by group, with the results of Independent Samples Mann-Whitney U testing illustrated in Figure 5.3.

The X-Bolt showed significantly less displacement in both stance and sitting tests ($p < 0.0001$).

Table 5.4 and Figure 5.4 illustrate this in the context of an intact, uninstrumented femur.

The X-Bolt device conferred significantly more resistance to torsion than the Affixus with a single compression screw in a 31A2 fracture model, when subjected to supra-physiological but proportionally balanced loads simulating a worst-case of those encountered during sitting and standing.

Table 5.1 – Mean displacement in stance and sitting tests by specimen

Group	Specimen	Mean stance displacement	Mean sitting displacement
X-Bolt	1	-10.38	6.56
X-Bolt	2	-12.37	6.86
X-Bolt	3	-13.29	7.34
X-Bolt	4	-9.33	6.30
X-Bolt	5	-10.85	7.02
X-Bolt	6	-7.37	5.27
X-Bolt	7	-10.00	9.78
X-Bolt	8	-8.96	8.50
Affixus	9	-10.45	6.40
Affixus	10	-19.13	10.70
Affixus	11	-13.21	9.23
Affixus	12	-24.10	12.53

Table 5.2 – Shapiro-Wilk tests for normality of distribution of stance and sitting displacement in X-Bolt and Affixus models

Test	Group	Statistic	df	p value
Standing	X-Bolt	.873	80	<0.0001
displacement	Affixus	.883	38	0.001
Sitting	X-Bolt	.927	80	<0.0001
displacement	Affixus	.938	38	0.035

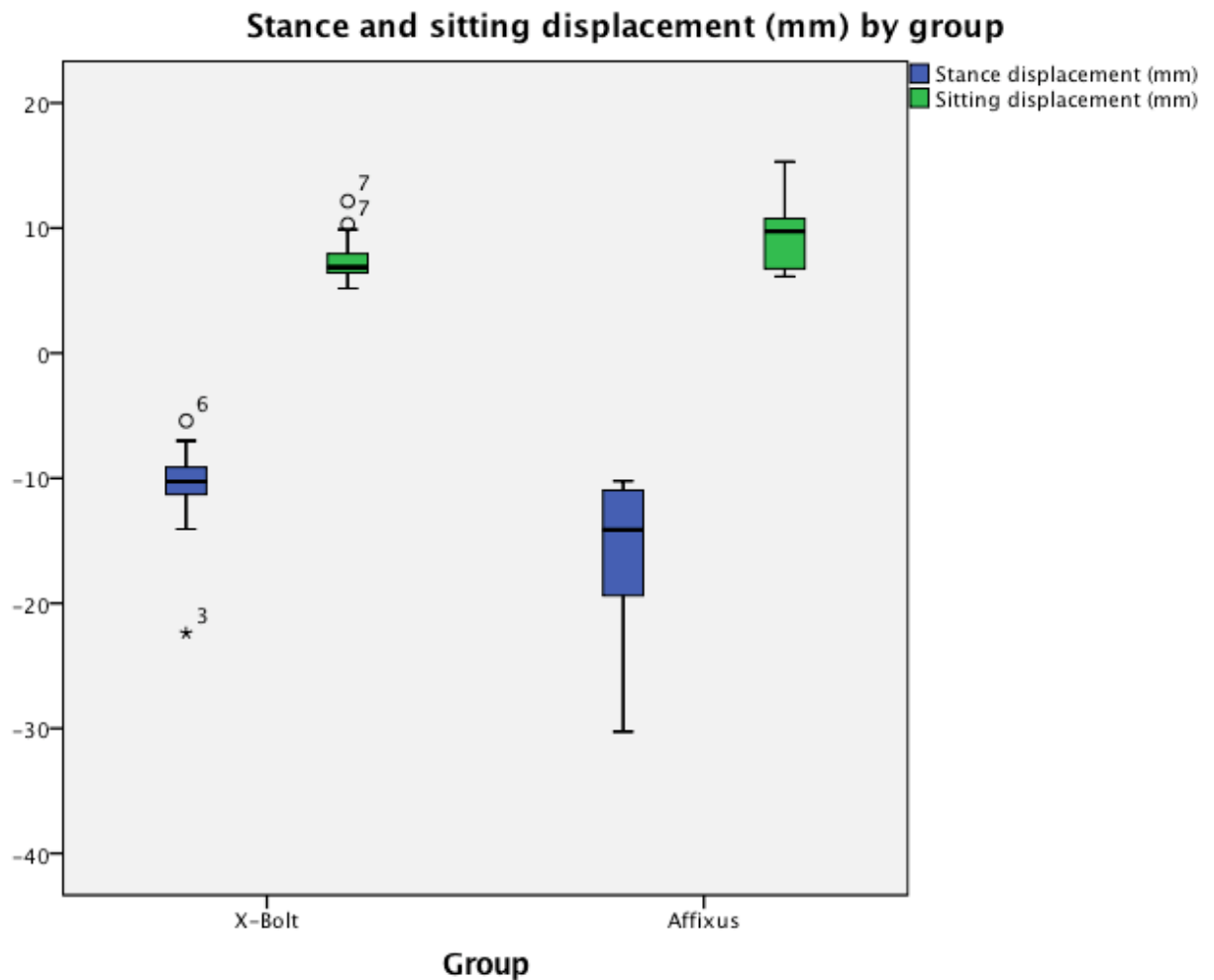


Figure 5.3 – Standing and sitting displacement by group, without intact femur (outliers fall outside the range -22.4 to -5.4 mm in stance or >10.3 mm sitting and are plotted as asymptotes)

Table 5.3 – Independent Samples Mann-Whitney U testing of standing and sitting displacement between X-Bolt and Affixus models

Group	Mean stance displacement (mm)	SD	p value	Mean sitting displacement (mm)	SD	p value
X-Bolt	-10.32	2.27	<0.0001	7.20	1.36	<0.0001
Affixus	-16.33	5.56		9.57	2.38	

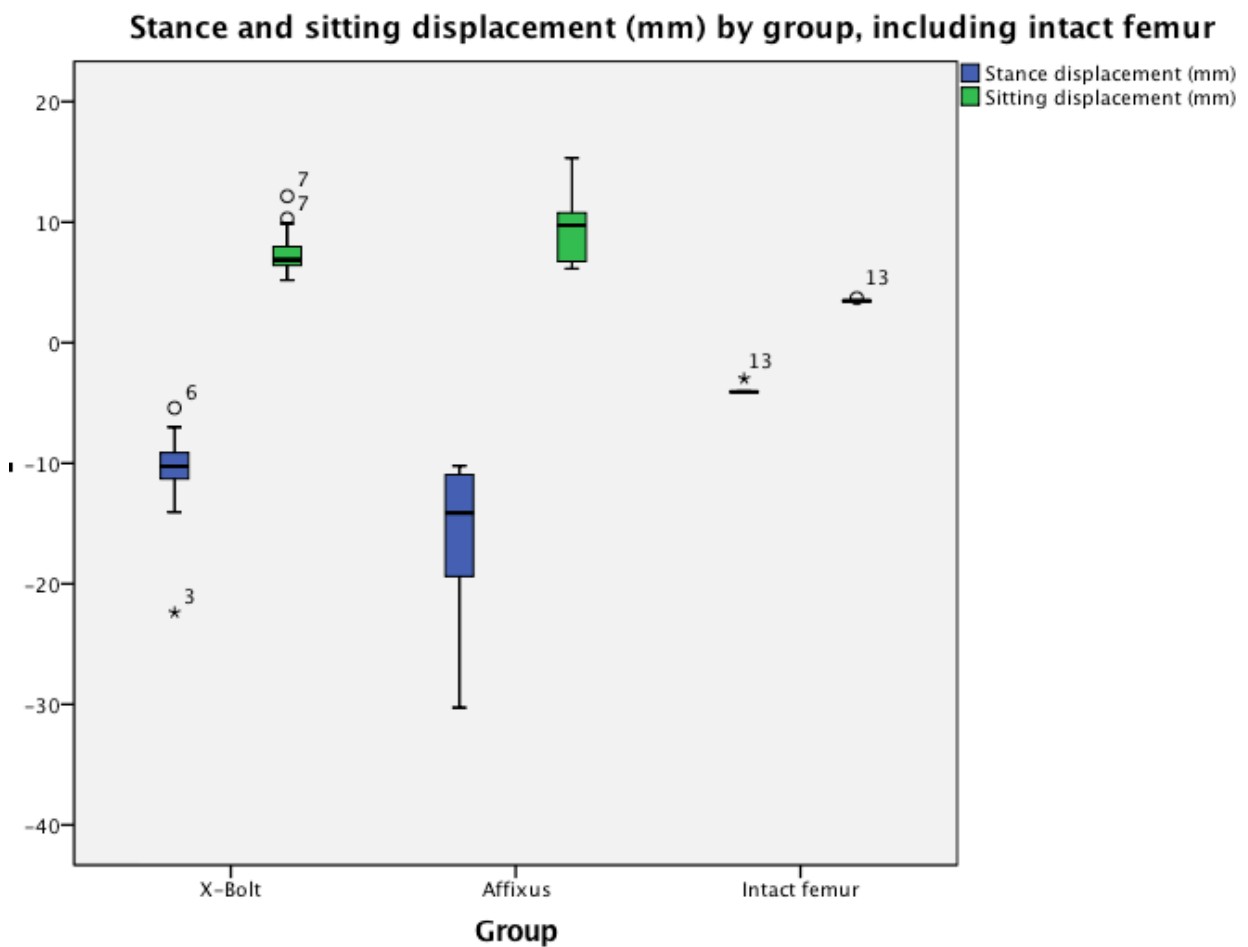


Figure 5.4 – Standing and sitting displacement by group, with un-instrumented femur (outliers fall outside the range -22.4 to -5.4 mm in stance or >10.3 mm sitting and are plotted as asymptotes)

Table 5.4 – Independent Samples Kruskal-Wallis testing of stance and sitting displacement between X-Bolt and Affixus models and un-instrumented femur

Group	Mean stance displacement (mm)	SD	p value	Mean sitting displacement (mm)	SD	p value
X-Bolt	-10.32	2.27	<0.0001	7.20	1.36	<0.0001
Affixus	-16.33	5.56		9.57	2.38	
Intact femur	-3.97	.35		3.49	.080	

5.5 Discussion

This experiment demonstrated superior resistance to torsion in the model using the X-Bolt. It further provided demonstration of the feasibility of using this method to exert the torsional force required to perform the testing. The novel technique for inducing torsional load was reproducible between specimens and failed in only one test, with the failure being of the model rather than the jig or testing apparatus. The displacement seen in this technique does not necessarily represent the resting state of the implant when all load is withdrawn, and it may be interesting to using a technique such as micro CT to ascertain how much of this deformation is permanent.

The magnitude and direction of loading was one based on worst-case, as seen in telemetry-equipped implant studies, and designed to be broadly reflective of forces acting during movement between sitting and standing, or climbing stairs. Pragmatism is required when interpreting these results – this test is designed far more to show that it is conceptually safe to consider an early clinical trial in which these activities would be undertaken than to mimic these complex biomechanical interactions closely.

Unlike in the previous experiment, there remains a significant difference between either model and an intact femur. This is to be expected, as in this experiment the fracture was incorporated into the tested region of model, whereas previously the testing was of intact femoral diaphysis, whether or not instrumented.

The manufacturers of the Affixus nail offer a second compression screw as an option in the construct, but this test compared it in its basic configuration. It could be argued that this is not the optimum possible performance of the implant, but equally it is an approved mode of use and so this could be argued to have been a fair and valid test. There has not been implant-specific testing of the Z effect phenomenon in the Affixus device, so it would also be interesting to see if there was a genuine benefit to this screw configuration.

As the same model was used as that in Chapter 4, the same imbalance in number of specimens existed in this experiment, with 8 X-Bolt and 4 Affixus models tested and the same issues relating to sample size are, therefore, pertinent. The catastrophic failure of the twelfth model is likely to relate to the experimental technique and so, whilst it cannot be discounted, it should be considered in this

context when evaluating the efficacy of the device. With further resources, it would have been ideal to create further models for testing.

Chapter 6: Discussion

6.1 Key findings and their context

6.1.1 *Clinical study*

The absolute numbers of patients experiencing problems sufficient to prompt representation to medical care after fixation of an extracapsular hip fracture are not high, but the results presented in Chapter 3 indicate a population exists which is not captured by the NHFD. These patients nevertheless are experiencing suboptimal outcomes, with commonly reported problems of failure to remobilize, limp and pain.

The rates of failure seen in this study are in keeping with those published contemporaneously, including by members of the team conducting this research project (Page et al., 2016), in a previous study of a similar population in another centre. This gives face validity to the results but may also contradict the anecdotal assertions that there are many more people with failing fixations than we realize. In particular, the assumption often used to justify use of a particular construct, that many people experience significant failure of fixation but with no cut-out, does not seem to be demonstrated in this study. While this would be far better investigated radiographically, it can be argued that even a significant radiographic shortening or mal-union is of no real consequence if not causing a patient problems. This was exactly the situation in Reindl's study, where a radiographic parameter differed between implants but functional outcomes did not (Reindl, Harvey, Berry, & Rahme, 2015). Furthermore, as routine follow up radiographic examination is not performed in the hip fracture population, these radiographs were not available as part of routine clinical care and recalling patients for additional radiological examination could not be ethically justified for this study.

The trend towards better ASA and AMTS in the group that experience problems in comparison to the control group is a novel and interesting finding. Whilst one may expect patients with higher ASA grades, and hence poorer health, to experience problems with or failures of fixation, it may in fact be the case that their functional levels are lower and hence they either do not have opportunity to stress their fixation in the same way that a fit individual would. By the same token, those with lower AMTS may have problems communicating these problems. As the detection of problems in this study design hinged on patients reporting them

to their doctor or other healthcare professional, this sheds some light on why apparently healthier patients had more problems – it may simply have been the case that they were better able to communicate them. This is a novel finding and one worthy of further research, particularly when considered in the context of this often co-morbid and cognitively impaired population.

The tip-apex distances in this study are acceptable in both groups but are higher in the group experiencing problems. When considering that the patients in the problem cohort have worse ASA grades, there are a number of potentially contributory factors. Many forms of chronic ill health are themselves risk factors for altered BMD, or the medications used to treat them may be. A combination of reduced BMD and increased TAD may be one that tips the balance in these precarious patients – a high-quality prospective study combining BMD and TAD has yet to be undertaken. Nevertheless, this finding highlights that the optimal TAD is the smallest that can be achieved, rather than simply aiming to be within Baumgaertner's classically-described 25mm (Baumgaertner et al., 1995).

6.12 Biomechanical studies

The distribution of these problems between those patients undergoing fixation with SHS and with IMN is not significantly different. The AO classification of the fractures in those patients with problems is also evenly distributed. All except one of the 31A1 fractures were fixed with an SHS, and 24 of 27 A2 fractures. It is notable that in the case group, 7 of 11 31A3 fractures were fixed by SHS, a much higher proportion than the 3 of 13 patients in the group without problems. This may suggest either a failure to recognize the complex nature of the fracture pre-operatively, or a decision outwith the guidance of CG124, resulting in an increased failure rate of this construct.

A similar composite stiffness exists between femora instrumented with 10 mm stainless steel nails and 11 mm titanium nails. This, in turn, suggests that the common concern over inappropriate stiffness and the concomitant risk of periprosthetic fracture may not be based on the performance of contemporary intramedullary devices regardless of the material from which they are manufactured.

The novel X-Bolt device exhibited a superior resistance to displacement within the femoral head when subjected to supra-physiological forces acting in the AP direction, to simulate extreme loading when standing and sitting.

The instrumented femoral model was challenging to create using economically viable synthetic bone femora. There were two primary modes of failure when creating the model, periprosthetic fracture and cortical breach from reaming.

6.13 The paucity of evidence behind the introduction of orthopaedic implants for trauma

A striking finding at the outset of this process was the absence of high-quality evidence in the literature to support devices in common use and which may currently be held to be standard of care.

A number of clinical cohort studies exist for many implants forming part of this review but they tend to be retrospective, non-randomized and, most crucially, describe the early results of a device already being implanted into patients in clinical practice. This is, conceptually, worlds apart from a phase 3 clinical trial, where both basic science and human feasibility studies have generated sufficient data to permit an ethical, methodologically robust randomized clinical trial.

Perry et al. (Perry et al., 2014) explored the issues around trauma trials in an instructional review, outlining surgeon, patient and disease factors. A key obstacle to improving trauma research was held to be surgeon equipoise – the whole basis of the process of investigation is that a question exists as to which implant or intervention is superior, and hence when surgeons believe they already know the answer, they will be unwilling to randomize their patients and risk them being exposed to what they perceive to be an inferior intervention.

Musculoskeletal care across the UK has been subject to a process of scrutiny and harmonisation termed ‘Getting it right first time’, an initiative aimed largely at ensuring resourcing and commissioning decisions are evidence-based, and equally that the interventions performed by orthopaedic surgeons and the implants used are equally well-supported (Briggs, 2012). One facet of this process has been a call for certain centres to become specialist centres, who are permitted to use newer devices in an environment of enhanced monitoring, data collection and robust governance. Whilst much of the thrust of the work is towards elective orthopaedic workload, these principles could be applied as well to trauma care.

6.2 Strengths and limitations

6.2.1 *Clinical considerations*

The clinical phase of this study relied on patients with problems after fixation of a hip fracture re-presenting to our centres, rather than seeking help elsewhere. A small proportion of patients sustain these injuries whilst away from home, and hence will likely seek follow-up in their local units. It is unquestionable that these patients are likely lost to our follow-up in the study design we have employed, but this is unlikely to be a significant number. Of our local patients, there remains the possibility that they would seek help elsewhere. Without much more agile, national datasets, this is again hard to quantify, but the model of primary care commissioning and referral makes it most likely that these patients will be referred back to the surgeon who treated them, or certainly their centre if not the surgeon themselves. The Hospital Episodes Statistics (HES) system would offer such information, but patient-level data is appropriately protected within this and so this would be hard to achieve, an unjustifiable use of resources given the absence of existing evidence to suggest this pattern of care episodes exists in this population.

The measurements taken from radiographs and the classification of AO/OTA grade relied on the surgeons gathering data at each centre. This classification system has been shown to be acceptable if used without its subgroups, and to have a reasonable interobserver reliability (Pervez et al., 2002). On this basis, when coupled with a process of observer agreement, there should be reasonable confidence in these results. By the same token, the measurement of TAD is arguably somewhat easier on Picture Archiving and Communication System (PACS), when able to readily calibrate the radiographs against known measurements such as the core diameter of the compression screw of the SHS.

The inability to derive the Garden alignment index from fluoroscopic views is an important finding. Few modern studies attempt to review the quality of reduction at all, but for any researchers planning to in future, it is likely that the methodology will need to adapt to reflect this. Given the problem was likely to be due to radiation dose control and auto-contrast features of modern image intensifiers, it may be that a formal AP radiograph is required post-operatively to

permit the best visualization of the trabeculae. This was not possible in this retrospective convenience sample but should be given consideration by anyone planning prospective research in the future. It does also mean that this study cannot consider the quality of reduction.

The relatively higher proportion of patients with higher AMTS in the case group could represent an increased capability to communicate about their problems, and hence seek referral or follow-up for discomfort or subtle mobility problems, whereas those with lower AMTS may have been reliant on others noticing more obvious limps or changes in patterns of mobility. Equally, an infected wound would tend to declare itself by externally visible signs such as the leakage of pus. It may therefore be the case that less noticeable problems are under-represented in patients with poorer overall health and with cognitive and communication difficulties. This would effectively represent an intrinsic bias in the methodology. The number of patients in the subgroups in this study prohibit meaningful analysis of this, but it is a point worth considering for further research.

The study methodology precluded gathering accurate data on timelines of failure. The use of multiple data sources, clinic appointments made some time after receipt of a GP letter (itself likely to be written some time after first appearance of the problem for which the patient has presented) and absent timelines in documentation meant that overall no standardized data could be obtained in this retrospective context. It would have been useful information by which to inform a further study; it is likely, however, that wide-ranging and prospective studies such as the WHiTE embedded trials will be able to capture this far better.

The use of the “Best mobility” marker was a compromise to permit some assessment of mobility from differing data points, whilst endeavouring not to overstate the level of accuracy. It was also thought to be a pragmatic marker, reflecting the varying thresholds which may substantially alter quality and function of life, i.e. losing all mobility, losing the ability to leave the house, having to walk with significant assistance or losing fully independent mobility. The very occurrence of this situation reflects both the heterogeneity in data captured by centres and the subjectivity of the matter overall.

Missing data is a problem in any study, and in this context must be balanced with the validity of any means used by which to try and fill it. The “Best mobility” marker was such a solution for which some rationale could be shown but, when radiographs or NHFD data are missing, it is harder to bridge this gap. In this study, a decision was taken to avoid exclusion purely for missing data, as the primary outcome was incidence of problems and the aim to provide information on this. While the absence of data may then have precluded much more meaningful analysis, the very low incidence of problems overall made it important that these patients should still appear in the cohorts and hence contribute to this epidemiological picture. It has been argued that mandating case completeness for inclusion in epidemiological studies risks introducing bias as the missing data may not be random (Sterne et al., 2009). In the context of this study, missing data could well reflect those patients being most socially isolated and under-supported, hence insufficient information being available on their pre-injury state, or in radiographic terms a patient who is too sick to leave the ward for imaging in the radiology department. Overlooking either such group in an epidemiological study would, therefore, introduce clear bias.

Although not an inherent weakness in this study as such, a number of patients were found to have been mis-coded in the NHFD. While it is inevitable that such a large dataset will contain mistakes, a key feature was that they mainly seemed to relate to fracture classification and subsequent operation, in that some patients recorded as “Intertrochanteric” had in fact had SHS fixation for an undisplaced intracapsular fracture. This may then suggest a training issue to be addressed with those responsible for data processing or recording, to ensure they understand that the use of a SHS is not pathognomonic of intertrochanteric fracture.

6.22 Femoral model

The composite synthetic femur models used in this series of experiments were variants designed to mimic normal bone density, rather than the reduced state found in osteoporosis. This was due to a cost constraint, with only a significantly more expensive manufacturer offering an osteoporotic variant. As the models used were standardized across all experiments, a constant density was under investigation and so any effect should at least be universal. This approach does risk, however, that any device performing unexpectedly better or worse in reduced BMD would not necessarily be obvious in this study.

The difficulties in instrumenting the femora in the Affixus group were likely due to the increased diameter of the nail by contrast with the X-Bolt group. In clinical practice, patients with such high cortical thickness and minimal intramedullary canal space are rarely encountered in this group. It is likely that this model had a very high proportion of the nail in cortical contact; if an *in vivo* fixation operation was this challenging, something would be changed intra-operatively to avoid periprosthetic fracture. Typically, this step would either be to down-size the nail or to ream the femoral canal further. On this basis, it may be unlikely to encounter a patient with an IMN of such size with such cortical contact. There is also a difference in radius of curvature of the nails, with the X-Bolt nail being a more gentle one. If already challenged by the femoral anatomy in the standardized models, a more curved nail may have been the decisive factor in failure of the model. The resultant model in this experiment, if the IMN was over-sized in relation to the femur, should have been more prone to catastrophic failure and so the survival of the model under supra-physiological loads should be, if anything, more reassuring.

By the same token, the use of cadaveric material in place of synthetic bones may have provided more biofidelity, but the evidence for the generalizability of results obtained in synthetic bone to those one might expect in human bone, discussed earlier in this thesis, is reassuring. The cost, logistic and legal implications of Human Tissue Authority governance would render any requirement for cadaveric work exponentially more expensive than its synthetic bone equivalent. It may, however, be a necessary step to reassure surgeons and patients alike that the

implantation of this device is feasible, and its performance under these conditions adequate.

Basso et al. reported on a comparison of fourth-generation composite femora with cadaveric osteoporotic bones and held them not be comparable, with the composite femora failing with short oblique or transverse fractures in a way not seen in the cadaveric femora (Basso et al., 2014). This reflects some of the failures seen in these experiments, potentially supporting the suggestion that cadaveric testing may yield different results.

The unequal group sizes make this experiment vulnerable to error. The 3-point bending experiment found a femur instrumented with a steel nail not significantly stiffer than one instrumented with a titanium nail in these models, which could represent a type II error, while a type I error in the second experiment could have led to an incorrect assertion that the X-Bolt device is significantly more resistant to torsion than a conventional compression screw. Any experiments with more complex models to address the limitations of the sawbones constructs would likely benefit from larger groups. This series of experiments should, however give an idea of the magnitude of effects and hence better inform sample sizes for future work.

Recent work has demonstrated the feasibility of using acid-degraded bovine long bones in orthopaedic research, with the resultant sections having very similar biomechanical properties to human normal and osteoporotic bone. This technique would not, however, have been suitable for use in this study, which relied on the morphology of the human femur for instrumentation.

6.23 *Experimental methodology*

The forces used in the 3-point bending experiment were designed to exceed those described in previous work describing physiological forces acting across the femur during various activities of daily living (D'Angeli et al., 2013). The application of 500N in this experiment was intended to produce a worst-case scenario in excess of these but was only performed once. It could be argued that, especially in vivo, the effect of repeated loading may be more complex with a cycle of microfracture, propagation and eventual failure. It may be desirable, therefore, to perform a similar experiment with cyclical loading, using failure as an endpoint.

It is also possible that as the femoral head was not constrained by the acetabulum and its surrounding soft tissues in this experiment, more displacement of the head was possible when the model was bending, thus rendering the whole construct less stiff. In clinical practice, however, there ought to be little motion at the fracture site once it has collapsed, if it is well-reduced, and so the effect of this is likely to be negligible.

The motion used in the resistance to torsion experiment is not one which has been validated previously as a means of assessing intertrochanteric fixation devices. Lenich's work was premised on the risk of cavitation around an off-centre implant, and in the same way a propensity to permit substantial movement within the femoral head in the AP plane could confer the same risk (Lenich et al., 2011).

There exists, nonetheless, a great deal of evidence showing very high loads and contact pressures across and within the hip in the AP direction during sitting and standing. It is therefore reasonable to investigate them further, especially in the context of the X-Bolt device where the straightforward resistance to directly-applied rotational deformation is already known (Gosiewski et al., 2017). Little work has, however, been conducted on the more complex model of load in the AP direction whilst rotation is being applied.

Other authors have used more complex, polyaxial systems for testing devices with composite loading patterns. Santoni et al.'s cadaveric experiment to evaluate the rotational stability of the InterTAN system, for example, used a rotating arm and

load frame to subject the specimen to torsion in the axis around the femoral neck fixation and to the axial load simulating body weight (Santoni et al., 2016). In this way, a similar force was being applied in the AP direction, once the rotational element of the force had been applied. Such a test would have been an attractive option to use here, but fell outwith the constraints of the available equipment.

When testing a centrally-sited device within a spherical femoral head model, it could also be argued that the direction of force becomes less relevant in any case. It is desirable for a device to be as resistant as possible to cut-out in any plane, and hence in this experiment we have simply sought to make the forces clinically relevant.

There is a risk that this measurement reflects the degree of bending in the fixation device, rather than cavitation. Micro-CT may be a suitable modality for assessing this by scanning the models at the end of the cycle. Whether cavitation in the head or bending of the implant, however, the strain environment across a reduced fracture would be altered in either case and so the risk of disruption of union is likely to be increased regardless.

The forces and manner in which they were applied are designed to have as much bio-fidelity as possible, but they cannot replicate the complex, polyaxial forces exerted on any joint, nor the microscopic and very high-frequency neuromuscular inputs which mean that for something to stay still, there are generally many dynamic processes unfolding simultaneously. Biomechanical studies tend to all be affected by this problem, and it is a clear illustration of why neither basic science nor clinical studies alone can ever provide enough information to truly understand every aspect of an implant's performance.

This may be a useful area in which to deploy FEA, as discussed earlier, to permit more complex modelling of the forces acting through the joint and to iteratively improve the design of an implant as much as possible before taking the measured clinical risk of implantation into a patient. This may be a better approach than attempting increasingly complex biomechanical experiments which, while potentially expensive in laboratory time and consumables may still not come close to approximating conditions *in vivo*.

The cycles of loading applied during the biomechanical experiments were representative of the magnitude and vector of forces to which a fixation construct may be subject in activities of daily living, but not of the duration or repetition. This was based on the principle that maximum displacement would be most likely to occur when the force was first applied, with the result of repetition tending more towards failure through fatigue. Such failure was not an outcome under investigation in this study but is undoubtedly one that should form part of the pre-clinical assessments made before an implant can be considered for early trials.

6.3 Methodological lessons learned

In addition to the actual experimental results, this process has provided learning points when considering how to perform this type of research.

To test implants, the support of the companies designing, manufacturing and supplying them is essential. The very high cost of the implants, coupled with a very regulated distribution system means that devices can only be obtained through the company. Further, adherence to operative technique requires access to equipment-specific instrumentation, which can be usually be obtained on a free loan basis from the manufacturer, in a serviceable condition. This contrasts with the unregulated, online second-hand market, where high costs may be combined with completely unserviceable equipment. By using manufacturers already supplying local hospitals, it was easier to build on existing relationships and to make use of existing logistic chains.

Simulation of physiological loads, and especially the direction in which they are applied in a joint with many axes of movement is challenging. It can be argued that a law of diminishing returns may exist – even if forces could be exerted in many directions at once during testing, and applied in the values and proportion in which they act *in vivo*, it remains the case that biomechanical testing is by its nature a crude approximation. The setup of a more rudimentary experiment such as those performed in this study lacks the extrinsic stabilizers which would normally act on the joint and, even in cadaveric experiments, the static stabilizers such as ligaments may be preserved, but the dynamic stabilization provided by muscles is still lacking.

It is, therefore, clear that the multi-staged approach to testing medical interventions is vital for new surgical implants. Basic engineering and metallurgical tests are required to prove the design and viability of a device, and then biomechanical experimentation can prove the concept at a very experimental level.

For higher quality evidence, however, clinical trials are required. Prior to this, a limited cohort study should provide a translational bridge between the laboratory research and a randomized controlled trial. Such an approach minimizes risk by

limiting the number of patients exposed to the risk of an experimental intervention and, in surgical techniques, should also limit the number of clinicians performing the intervention. It also permits more detailed experimental strategies, such as the use of radiostereometric analysis (RSA) to assess migration, which may not be possible once embarked on a larger clinical trial. This concept was introduced more than 20 years ago by Gross, and further built upon by Malchau (Gross, 1993; Malchau et al., 2011). Despite this, metal on metal hip implants were introduced to the market amidst equivocal evidence and some known engineering concerns (Cohen, 2012). Incidents such as this, disregarding the processes advocated by surgeons such as Gross and Malchau, should clearly underline the inherent value in the stepwise introduction of devices. This has been espoused in the “Get it right first time” agenda, where the risk from the requirement for devices to be implanted in order to generate data is balanced by limiting the number of the centres permitted to do so, and subjecting them to increased surveillance (Briggs, 2012). In this way, any problems should be both detected at an earlier stage and more limited in incidence.

6.4 Conclusions

This study has demonstrated that the majority of patients who undergo fixation of hip fractures do not go on to have further problems. Of those who do, not all are captured by the NHFD and a number re-present to hip surgeons. Of those who do, this study suggests that subtler problems are only likely to be brought to medical attention if the patient is in better health and able to communicate. There is an over-representation of 31A3 fractures fixed by SHS in the cohort of patients with problems, reinforcing existing evidence for the use of IMN in this subgroup and hence offering a supportive rationale for the ongoing evolution and development of this class of device.

The study has also demonstrated that stainless steel nails can perform similarly to titanium ones in respect of stiffness in the construct of an instrumented femur, with no significant difference seen. This suggests that there should not be a reflex unwillingness to consider such implants manufactured from stainless steel – but equally the good design principles seen in this device should be respected when considering this evidence in the design of other devices. These include, proximal and distal diameter, tapering and fluting of the device.

The X-Bolt novel expanding bolt has been shown to have superior resistance to displacement induced by torsion, in the manner in which it might be applied moving between sitting and standing positions, or when climbing stairs. The methodology employed for this testing was itself novel and, given its relative simplicity, should be considered for future testing of any hip fracture fixation device. In an injury where the primary mode of failure remains cut-out of the compression device, such a potential benefit warrants further investigation. The absence of any failures of the device at significantly supra-physiological loads should be reassuring for those considering early clinical trials of the device.

Overall, this thesis has demonstrated that failure of fixation occurs in extracapsular hip fractures, and that a preponderance of this is seen in those less stable A3 fractures fixed with sliding hip screws. This reinforces the concept that there is a place for intramedullary devices in the fixation of trochanteric hip fractures, where guidelines already recommend the intervention. The fact that some failures of intramedullary fixation occur suggests that room remains for a

device which performs better. It may be that this cannot be found, that the limits of engineering have been reached and we should focus instead on patient optimisation, rehabilitation and acceptance that perfect surgical outcomes are unattainable. This notwithstanding, the potential benefits in improved outcome extrapolated to a large population could well justify progression to at least a feasibility trial. The project has shown that the X-Bolt nail is safe in terms of its metallurgy and design, and that the expanding bolt element of it may offer benefit over standard of care devices. This project should serve as evidence supportive of progression of testing to early clinical trials, which should be undertaken in a progressive and cautious fashion.

6.5 Future research directions

The most obvious research direction from this work is a clinical trial to compare these implants in the nail configuration. The X-Bolt device itself is now a veteran of several large, robust clinical trials, but is as yet unevaluated when used in conjunction with an intramedullary nail rather than a plate on the lateral aspect of the femur. The data from the clinical phase of this study, with a problem rate of around 5% and failure rate of a third of these suggests that a trial endpoint of failure of the device in a non-inferiority design would need nearly 7000 patients (6718 patients with 90% power, 5% alpha, 98% success rate in each arm and 1% non-inferiority margin). Continuing the trend away from surgical endpoints and towards patient benefit and experience would suggest such a trial could make use of the core outcomes already defined as part of the WHiTE programme of embedded trials (Haywood et al., 2014).

As discussed in chapter 5.3, however, such a trial should come only after appropriate limited clinical and feasibility studies. The X-Bolt device has been evaluated and has not raised any concerns in its performance in a large multicentre RCT. The 3-point bending evidence from this study should provide reassurance that implantation is safe; if anything, the model here should be theoretically more prone to periprosthetic fracture than the femur of a living patient. It may, therefore, now be appropriate to proceed to a limited cohort study. Such a study, with both radiological outcomes derived from a modality such as RSA and functional and quality of life outcomes, could give confidence that a full-scale RCT would be both feasible and ethical.

A realistic appraisal of the current climate around surgical device testing, especially in the context of catastrophic failure of metal on metal total hip replacements and, very recently, the meshes used for urogynaecological reconstruction suggests that an interim step may be required. This may be best achieved by a repetition of these experiments in cadaveric femora, providing reassurance that the performance of the implant in human bone is not dissimilar to its performance in relatively high fidelity synthetic femora.

Whether or not cadaveric work is needed, a tapered introduction to use in humans is essential. The support from the manufacturer of the device for a number of

trials, both ongoing and already concluded, and the lessons learned from recent adverse launches to market could now be combined to make the orthopaedic surgical profession an exemplar of best practice in the introduction of a new device.

With the eventual completion of such an RCT, the full journey of orthopaedic innovation described by Gross (Gross, 1993) would be travelled, bringing a safe implant to market with a full body of supporting evidence.

Bibliography

Al-Ashqar, M., Panteli, M., Chakrabarty, G. & Giannoudis, P.V. (2018) Atypical fractures: An issue of concern or a myth? *Injury*. 49 (3), 649–655.

Andreani, L., Bonicoli, E., Piolanti, N., Parchi, P., Niccolai, F., Carmignani, A. & Lisanti, M. (2015) Prospective Randomized Controlled Trial of Two Different Intramedullary Nails for Pertrochanteric Fractures of the Femur. *Surgical Technology International*. 27210–214.

Arens, S., Schlegel, U., Printzen, G., Ziegler, W.J., Perren, S.M. & Hansis, M. (1996) Influence of materials for fixation implants on local infection. An experimental study of steel versus titanium DCP in rabbits. *The Journal of Bone and Joint Surgery. British volume*. 78 (4), 647–651.

Augat, P., Simon, U., Liedert, A. & Claes, L. (2004) Mechanics and mechanobiology of fracture healing in normal and osteoporotic bone. *Osteoporosis International*. 16 (S02), S36–S43.

Babst, R., Renner, N., Biedermann, M., Rosso, R., Heberer, M., Harder, F. & Regazzoni, P. (1998) Clinical results using the trochanter stabilizing plate (TSP): the modular extension of the dynamic hip screw (DHS) for internal fixation of selected unstable intertrochanteric fractures. *Journal of Orthopaedic Trauma*. 12 (6), 392–399.

Bartoniček, J. & Rammelt, S. (2014) The history of internal fixation of proximal femur fractures Ernst Pohl—the genius behind. *International Orthopaedics*. 38 (11), 2421–2426.

Basso, T., Klaksvik, J., Syversen, U. & Foss, O.A. (2014) A biomechanical comparison of composite femurs and cadaver femurs used in experiments on operated hip fractures. *Journal of Biomechanics*. 47 (16), 3898–3902.

- Basso, T., Klaksvik, J., Syversen, U. & Foss, O.A. (2012) Biomechanical femoral neck fracture experiments--a narrative review. *Injury*. 43 (10), 1633–1639.
- Baumgaertner, M.R., Curtin, S.L., Lindskog, D.M. & Keggi, J.M. (1995) The value of the tip-apex distance in predicting failure of fixation of peritrochanteric fractures of the hip. *The Journal of Bone and Joint Surgery*. 77 (7), 1058–1064.
- Berger-Groch, J., Rupprecht, M., Schoepper, S., Schroeder, M., Rueger, J.M. & Hoffmann, M. (2016) Five-Year Outcome Analysis of Intertrochanteric Femur Fractures: A Prospective Randomized Trial Comparing a 2-Screw and a Single-Screw Cephalomedullary Nail. *Journal of Orthopaedic Trauma*. 30 (9), 483–488.
- Bergmann, G., Bender, A., Dymke, J. & Duda, G. (2017) *Orthoload* [online]. Available from: www.orthoload.com (Accessed 1 March 2017).
- Bergmann, G., Bender, A., Dymke, J.R., Duda, G. & Damm, P. (2016) Standardized Loads Acting in Hip Implants M A P rez (ed.). *PloS One*. 11 (5), e0155612–23.
- Bergmann, G., Deuretzbacher, G., Heller, M., Graichen, F., Rohlmann, A., Strauss, J. & Duda, G.N. (2001) Hip contact forces and gait patterns from routine activities. *Journal of Biomechanics*. 34 (7), 859–871.
- Bergmann, G., Graichen, F. & Rohlmann, A. (1993) Hip joint loading during walking and running, measured in two patients. *Journal of Biomechanics*. 26 (8), 969–990.
- Bergmann, G., Graichen, F., Rohlmann, A., Bender, A., Heinlein, B., Duda, G.N., Heller, M.O. & Morlock, M.M. (2010) Realistic loads for testing hip implants. *Bio-medical Materials and Engineering*. 20 (2), 65–75.
- Blair, B., Koval, K.J., Kummer, F. & Zuckerman, J.D. (1994) Basicervical fractures of the proximal femur. A biomechanical study of 3 internal fixation techniques. *Clinical Orthopaedics and Related Research*. (306), 256–263.
- Bonnevialle, P., Saragaglia, D., Ehlinger, M., Tonetti, J., Maise, N., Adam, P. & Le Gall, C. (2011) Trochanteric locking nail versus arthroplasty in unstable

intertrochanteric fracture in patients aged over 75 years. *Orthopaedics & Traumatology: Surgery & Research*. 97 (6), S95–S100.

Boone, C., Carlberg, K.N., Koueiter, D.M., Baker, K.C., Sadowski, J., Wiater, P.J., Nowinski, G.P. & Grant, K. (2013) Short Versus Long Intramedullary Nails for Treatment of Intertrochanteric Femur Fractures (OTA 31-A1,A2). *Journal of Orthopaedic Trauma*. 28 (5), e96–e100.

Boutroy, S., Khosla, S., Sornay-Rendu, E., Zanchetta, M.B., McMahon, D.J., Zhang, C.A., Chapurlat, R.D., Zanchetta, J., Stein, E.M., Bogado, C., Majumdar, S., Burghardt, A.J. & Shane, E. (2016) Microarchitecture and Peripheral BMD are Impaired in Postmenopausal White Women With Fracture Independently of Total Hip T-Score: An International Multicenter Study. *Journal of Bone and Mineral Research*. 31 (6), 1158–1166.

Bridle, S.H., Patel, A.D., Bircher, M. & Calvert, P.T. (1991) Fixation of intertrochanteric fractures of the femur. A randomised prospective comparison of the gamma nail and the dynamic hip screw. *The Journal of Bone and Joint Surgery. British volume*. 73 (2), 330–334.

Briggs, T.W.R. (2012) Getting it right first time. *Improving the quality of orthopaedic care within the National Health Service in England*.

Bucholz, R.W., Ross, S.E., Lawrence, K.L.others (1987) Fatigue fracture of the interlocking nail in the treatment of fractures of the distal part of the femoral shaft. *The Journal of Bone and Joint Surgery. American volume*. 69 (9), 1391–1399.

Buckwalter, J., Glimcher, M., Cooper, R. & Recker, R. (2010) Bone Biology. *Journal of Bone and Joint Surgery*. 771256–1275.

Byrne, D.P., Mulhall, K.J. & Baker, J.F. (2010) Anatomy & Biomechanics of the Hip. *The Open Sports Medicine Journal*. 4 (1), 51–57.

Cai, L., Wang, T., Di, L., Hu, W. & Wang, J. (2016) Comparison of intramedullary and extramedullary fixation of stable intertrochanteric fractures in the elderly: a prospective randomised controlled trial exploring hidden perioperative blood loss. *BMC Musculoskeletal Disorders*. 17 (1), 475.

- Clawson, D.K. (1964) Trochanteric Fractures Treated by the Sliding Screw Plate Fixation Method. *Journal of Trauma and Acute Care Surgery*. 4 (6), 737–752.
- Colic, K., Sedmak, A., Grbovic, A., Tatic, U., Sedmak, S. & Djordjevic, B. (2016) Finite Element Modeling of Hip Implant Static Loading. *Procedia Engineering*. 149257–262.
- Cooper, A. (1822) *A treatise on dislocations and fractures of the joints*. 1st edition. London:.
- Cosman, F., de Beur, S.J., LeBoff, M.S., Lewiecki, E.M., Tanner, B., Randall, S. & Lindsay, R. (2014) Clinician's Guide to Prevention and Treatment of Osteoporosis. *Osteoporosis International*. 25 (10), 2359–2381.
- Cristofolini, L., Viceconti, M., Cappello, A. & Toni, A. (1996) Mechanical validation of whole bone composite femur models. *Journal of Biomechanics*. 29 (4), 525–535.
- Cummings, S.R. & Melton, L.J. (2002) Epidemiology and outcomes of osteoporotic fractures. *The Lancet*. 359 (9319), 1761–1767.
- Cummings, S.R., Browner, W.S., Bauer, D., Stone, K., Ensrud, K., Jamal, S. & Ettinger, B. (1998) Endogenous hormones and the risk of hip and vertebral fractures among older women. Study of Osteoporotic Fractures Research Group. *New England Journal of Medicine*. 339 (11), 733–738.
- Curtis, M.J., Jinnah, R.H., Wilson, V. & Cunningham, B.W. (1994) Proximal femoral fractures: a biomechanical study to compare intramedullary and extramedullary fixation. *Injury*. 25 (2), 99–104.
- Dargent-Molina, P., Favier, F., Grandjean, H., Baudoin, C., Schott, A.M., Hausherr, E., Meunier, P.J. & Breart, G. (1996) Fall-related factors and risk of hip fracture: the EPIDOS prospective study. *The Lancet*. 348 (9021), 145–149.
- Davison, J.N., Calder, S.J., Anderson, G.H., Ward, G., Jagger, C., Harper, W.M. & Gregg, P.J. (2001) Treatment for displaced intracapsular fracture of the proximal femur. A prospective, randomised trial in patients aged 65 to 79 years. *The Journal of Bone and Joint Surgery. British volume*. 83 (2), 206–212.

Dequeker, J., Johnell, O., Dilsen, G., Gennari, C., Vaz, A.L., Lyritis, G., Mazzuoli, G.D., Miravet, L., Passeri, M., Pérez-Cano, R. others (1993) Osteoarthritis protects against femoral neck fracture: the MEDOS study experience. *Bone*. 1451–56.

Desteli, E.E., İmren, Y., Erdoğan, M. & Aydagün, Ö. (2015) Quality of Life Following Treatment of Trochanteric Fractures with Proximal Femoral Nail versus Cementless Bipolar Hemiarthroplasty in Elderly. *Clinical and investigative medicine. Medecine clinique et experimentale*. 38 (2), E63–72.

Disegi, J.A. & Eschbach, L. (2000) Stainless steel in bone surgery. *Injury*. 31 Suppl 42–6.

Dix, D.B., Araoye, I.B., Staggers, J.R., Lin, C.P., Shah, A.B., Agarwal, A.K. & Naranje, S.M. (2018) A systematic review and meta-analysis of complications in conversion arthroplasty methods for failed intertrochanteric fracture fixation. *Journal of Clinical Orthopaedics and Trauma*. 1–4.

Drake, M.T., Clarke, B.L. & Khosla, S. (2008) Bisphosphonates: mechanism of action and role in clinical practice. *Mayo Clinic Proceedings*. 83 (9), 1032–1045.

Duriez, P., Devaux, T., Chantelot, C., Baudrier, N., Hery, J.Y., Mainard, D., Favier, T. & Massin, P. (2016) Is arthroplasty preferable to internal fixation for the treatment of extracapsular fracture of the upper femur in the elderly? *Orthopaedics & Traumatology: Surgery & Research*. 102 (6), 689–694.

Eberle, S., Gerber, C., Oldenburg, von, G., Hungerer, S. & Augat, P. (2009) Type of Hip Fracture Determines Load Share in Intramedullary Osteosynthesis. *Clinical Orthopaedics and Related Research*. 467 (8), 1972–1980.

Ehmke, L., Fitzpatrick, D., Krieg, J., Madey, S. & Bottlang, M. (2005) Lag screws for hip fracture fixation: Evaluation of migration resistance under simulated walking. *Journal of Orthopaedic Research*. 23 (6), 1329–1335.

Erduran, M., Karakaşlı, A., Ertem, F., Taylan, O., Yildiz, D.V., Çelik, S. & Havıtcıoğlu, H. (2011) Biomechanical effects of the distance from the fracture zone to the interlocking fixation screw of intramedullary nail. *Journal of Biomechanics*. 444–1.

- Evans, E.M. (1949) The treatment of trochanteric fractures of the femur. *The Journal of Bone and Joint Surgery. British volume*. 31B (2), 190–203.
- Eveleigh, R.J. (1995) A review of biomechanical studies of intramedullary nails. *Medical Engineering and Physics*. 17 (5), 323–331.
- Faresjö, T. & Faresjö, Å. (2010) To Match or Not to Match in Epidemiological Studies—Same Outcome but Less Power. *International Journal of Environmental Research and Public Health*. 7 (1), 325–332.
- Fletcher, J., Whitehouse, M., Gill, H. & Preatoni, E. (2017) 'Establishing and validating an osteoporotic model using bovine tibia', in [Online]. July 2017
- Gabriel, S.B., Panaino, J.V.P., Santos, I.D., Araujo, L.S., Mei, P.R., de Almeida, L.H. & Nunes, C.A. (2012) Characterization of a new beta titanium alloy, Ti–12Mo–3Nb, for biomedical applications. *Journal of Alloys and Compounds*. 536S208–S210.
- Garden, R.S. (1971) Malreduction and avascular necrosis in subcapital fractures of the femur. *The Journal of Bone and Joint Surgery. British volume*. 53 (2), 183–197.
- Garden, R.S. (1964) Stability and union in subcapital fractures of the femur. *The Journal of Bone and Joint Surgery. British volume*. 46630–647.
- Gibson, D., Keogh, C. & Morris, S. (2012) A biomechanical study comparing the dynamic hip screw with an X-Bolt in an unstable intertrochanteric fracture model of the proximal femur. *J Bone Joint Surg [Br]*. 94 (Suppl XXXIX), 164.
- Gjertsen, J.-E., Fevang, J.M., Matre, K., Vinje, T. & Engesæter, L.B. (2011) Clinical outcome after undisplaced femoral neck fractures. *Acta Orthopaedica*. 82 (3), 268–274.
- Goffin, J.M., Pankaj, P. & Simpson, A.H. (2012) The importance of lag screw position for the stabilization of trochanteric fractures with a sliding hip screw: A subject-specific finite element study. *Journal of Orthopaedic Research*. 31 (4), 596–600.

- Gosiewski, J.D., Holsgrove, T.P. & Gill, H.S. (2017) The efficacy of rotational control designs in promoting torsional stability of hip fracture fixation. *Bone and Joint Research*. 6 (5), 270–276.
- Griffin, X.L., McArthur, J., Achten, J., Parsons, N. & Costa, M.L. (2013) The Warwick Hip Trauma Evaluation One -an abridged protocol for the WHiTE One Study: An embedded randomised trial comparing the X-bolt with slidinghip screw fixation in extracapsular hip fractures. *Bone and Joint Research*. 2 (10), 206–209.
- Griffin, X.L., Parsons, N., Achten, J., Fernandez, M. & Costa, M.L. (2015) Recovery of health-related quality of life in a United Kingdom hip fracture population. The Warwick Hip Trauma Evaluation--a prospective cohort study. *The Bone & Joint Journal*. 97-B (3), 372–382.
- Griffin, X.L., Parsons, N., McArthur, J. & Achten, J. (2016) The Warwick Hip Trauma Evaluation One. *Bone Joint*
- Gross, M. (1993) Innovations in surgery. A proposal for phased clinical trials. *The Journal of Bone and Joint Surgery. British volume*. 75 (3), 351–354.
- Gundle, R., Gargan, M.F. & Simpson, A.H. (1995) How to minimize failures of fixation of unstable intertrochanteric fractures. *Injury*. 26 (9), 611–614.
- Halder, S.C. (1992) The Gamma nail for peritrochanteric fractures. *Journal of Bone and Joint Surgery-British Volume*. 74 (3), 340–344.
- Hawley, G.W. (1938) *Bone setting*. (US2133859A).
- Hawley, G.W. & Padula, R.D. (1938) A bone plate which will not break or bend. *The Journal of Bone and Joint Surgery*. 20 (2), 469.
- Haynes, R.C., Pöll, R.G., Miles, A.W. & Weston, R.B. (1997) Failure of femoral head fixation: a cadaveric analysis of lag screw cut-out with the gamma locking nail and AO dynamic hip screw. *Injury*. 28 (5-6), 337–341.
- Heiner, A.D. (2008) Structural properties of fourth-generation composite femurs and tibias. *Journal of Biomechanics*. 41 (15), 3282–3284.

- Heiner, A.D. & Brown, T.D. (2001) Structural properties of a new design of composite replicate femurs and tibias. *Journal of Biomechanics*. 34 (6), 773–781.
- Helwig, P., Faust, G., Hindenlang, U., Hirschmüller, A., Konstantinidis, L., Bahrs, C., Südkamp, N. & Schneider, R. (2009) Finite element analysis of four different implants inserted in different positions to stabilize an idealized trochanteric femoral fracture. *Injury*. 40 (3), 288–295.
- Henley, M.B., Meier, M. & Tencer, A.F. (1993) Influences of Some Design Parameters on the Biomechanics of the Unreamed Tibial Intramedullary Nail. *Journal of Orthopaedic Trauma*. 7 (4).
- Hernlund, E., Svedbom, A., Ivergård, M., Compston, J., Cooper, C., Stenmark, J., McCloskey, E.V., Jönsson, B. & Kanis, J.A. (2013) Osteoporosis in the European Union: medical management, epidemiology and economic burden. *Archives of Osteoporosis*. 8 (1-2), 136.
- Heyse-Moore, G.H., MacEachern, A.G. & Evans, D.C. (1983) Treatment of intertrochanteric fractures of the femur. A comparison of the Richards screw-plate with the Jewett nail-plate. *The Journal of Bone and Joint Surgery. British volume*. 65 (3), 262–267.
- Hopp, S., Wirbel, R., Ojodu, I., Pizanis, A., Pohlemann, T. & Fleischer, J. (2016) Does the implant make the difference ? - Prospective comparison of two different proximal femur nails. *Acta Orthopaedica Belgica*. 82 (2), 319–331.
- Huiskes, R. & Bio, E.C.J. (1983) A survey of finite element methods in biomechanics. *Journal of Biomechanics*. 16 (6), 385–409.
- Jacob, J., Desai, A. & Trompeter, A. (2017) Decision Making in the Management of Extracapsular Fractures of the Proximal Femur – is the Dynamic Hip Screw the Prevailing Gold Standard? *The Open Orthopaedics Journal*. 11 (Suppl-7, M5), 1213–1217.
- Jacobs, R.R., Armstrong, H.J., Whitaker, J.H. & Pazell, J. (1976) Treatment of intertrochanteric hip fractures with a compression hip screw and a nail plate. *The Journal of Trauma: Injury, Infection, and Critical Care*. 16 (08), 599–603.

- Järvinen, T.L.N., Sievänen, H., Khan, K.M., Heinonen, A. & Kannus, P. (2008) Shifting the focus in fracture prevention from osteoporosis to falls. *BMJ*. 336 (7636), 124–126.
- Jensen, J.S. & Tøndevold, E. (1979) Mortality after hip fractures. *Acta Orthopaedica Scandinavica*.
- Jewett, E.L. (1941) One-piece angle nail for trochanteric fractures. *The Journal of Bone and Joint Surgery. American volume*.
- Johansson, S. (2009) On the Operative Treatment of Medial Fractures of the Neck of the Femur. *Acta Orthopaedica Scandinavica*. 3 (3-4), 362–392.
- Kammerlander, C., Pfeufer, D., Lisitano, L.A., Mehaffey, S., Böcker, W. & Neuerburg, C. (2018) Inability of Older Adult Patients with Hip Fracture to Maintain Postoperative Weight-Bearing Restrictions. *The Journal of Bone and Joint Surgery*. 100 (11), 936–941.
- Kanakaris, N.K., Tosounidis, T.H. & Giannoudis, P.V. (2015) Nailing intertrochanteric hip fractures: short versus long; locked versus nonlocked. *Journal of Orthopaedic Trauma*. 29 Suppl 4S10–6.
- Kanis, J.A. (2002) Diagnosis of osteoporosis and assessment of fracture risk. *The Lancet*. 359 (9321), 1929–1936.
- Kanis, J.A., Melton, L.J., Christiansen, C., Johnston, C.C. & Khaltaev, N. (1994) The diagnosis of osteoporosis. *Journal of Bone and Mineral Research*. 9 (8), 1137–1141.
- Keene, G.S., Parker, M.J. & Pryor, G.A. (1993) Mortality and morbidity after hip fractures. *BMJ*. 307 (6914), 1248–1250.
- Kellam, J.F., Meinberg, E.G., Agel, J., Karam, M.D. & Roberts, C.S. (2018) Introduction: Fracture and Dislocation Classification Compendium-2018: International Comprehensive Classification of Fractures and Dislocations Committee. *Journal of Orthopaedic Trauma*. 32 Suppl 1S1–S10.

- Kuroda, D., Niinomii, M., Morinaga, M., Kato, Y. & Yashiro, T. (1998) Design and mechanical properties of new. *Materials Science and Engineering*. 244–249.
- Kyle, R.F., Schaffhausen, J.M. & Bechtold, J.E. (1991) Biomechanical characteristics of interlocking femoral nails in the treatment of complex femoral fractures. *Clinical Orthopaedics and Related Research*. (267), 169–173.
- Leal, J., Gray, A.M., Prieto-Alhambra, D., Arden, N.K., Cooper, C., Javaid, M.K., Judge, A. REFRESH study group (2015) Impact of hip fracture on hospital care costs: a population-based study. *Osteoporosis International*. 27 (2), 549–558.
- Lenich, A., Bachmeier, S., Prantl, L., Nerlich, M., Hammer, J., Mayr, E., Al-Munajjed, A.A. & Füchtmeier, B. (2011) Is the rotation of the femoral head a potential initiation for cutting out? A theoretical and experimental approach. *BMC Musculoskeletal Disorders*. 12 (1), 79.
- Leung, K.S., So, W.S., Shen, W.Y. & Hui, P.W. (1992) Gamma nails and dynamic hip screws for peritrochanteric fractures. A randomised prospective study in elderly patients. *The Journal of Bone and Joint Surgery. British volume*. 74 (3), 345–351.
- Li, X., Zhang, L., Hou, Z., Meng, Z., Chen, W., Wang, P. & Zhang, Y. (2015) Distal locked and unlocked nailing for perthrochanteric fractures--a prospective comparative randomized study. *International Orthopaedics*. 39 (8), 1645–1652.
- Loizou, C.L. & Parker, M.J. (2009) Avascular necrosis after internal fixation of intracapsular hip fractures; a study of the outcome for 1023 patients. *Injury*. 40 (11), 1143–1146.
- Lorentzon, M. & Cummings, S.R. (2015) Osteoporosis: the evolution of a diagnosis. *Journal of Internal Medicine*. 277 (6), 650–661.
- Lustenberger, A., Bekic, J. & Ganz, R. (1995) [Rotational instability of trochanteric femoral fractures secured with the dynamic hip screw. A radiologic analysis]. *Der Unfallchirurg*. 98 (10), 514–517.
- Ma, J.-X., Kuang, M.-J., Fan, Z.-R., Xing, F., Zhao, Y.-L., Zhang, L.-K., Chen, H.-T., Han, C. & Ma, X.-L. (2017) Comparison of clinical outcomes with InterTan vs

- Gamma nail or PFNA in the treatment of intertrochanteric fractures: A meta-analysis. *Nature Scientific Reports*. 7 (1), 1–9.
- Maestro, A. (2013) Hip Fractures with Rotational Instability: Concept and Surgical Technique. *Surgery*. 03 (04), 1–4.
- Marsh, D. (2017) How to start a world class audit from scratch. NHFD 10th Anniversary Conference
- Marsh, J.L. (2002) Removal of radiation dose response effects: an example of over-matching. *BMJ*. 325 (7359), 327–330.
- Massoud, E.I.E. (2009) Fixation of basicervical and related fractures. *International Orthopaedics*. 34 (4), 577–582.
- Morris, D.S., Rohrbach, J.N., Rogers, M., Sundaram, L.M.T., Sonnad, S., Pascual, J., Sarani, B., D, P.R.M. & Sims, C. (2011) The Surgical Revolving Door: Risk Factors for Hospital Readmission. *Journal of Surgical Research*. 1–5.
- Müller, M.E., Nazarian, S., Koch, P. & Schatzker, J. (2012) *The comprehensive classification of fractures of long bones*. Springer Science & Business Media.
- National Clinical Guideline Centre (2017) *The management of hip fracture in adults*. 1–664.
- National Insitute for Health and Care Excellence (2017) *Hip fracture: management*. 1–19.
- Nedza, S.M., Fry, D.E., DesHarnais, S., Spencer, E. & Yep, P. (2017) Emergency Department Visits Following Joint Replacement Surgery in an Era of Mandatory Bundled Payments Lowell W Gerson (ed.). *Academic Emergency Medicine*. 24 (2), 236–245.
- Oakley, B., Nightingale, J., Moran, C.G. & Moppett, I.K. (2017) Does achieving the best practice tariff improve outcomes in hip fracture patients? An observational cohort study. *BMJ Open*. 7 (2), e014190–8.

Oehler, W.D. & Janka, P. (1983) [Osteosynthesis of unstable pertrochanteric femoral fractures with the compression hip screw]. *Aktuelle Traumatologie*. 13 (4), 172–174.

Orthopaedic Trauma Association Classification, Database and Outcomes Committee (2007) **Fracture and Dislocation Classification Compendium - 2007**

. *Journal of Orthopaedic Trauma*. 21 (S10), 1–12.

OTA Classification, Outcomes and Database Committee (2018) The OTA Fracture Classification Compendium - Femur. *Journal of Orthopaedic Trauma*. 32 Suppl 1S33–S44.

Ozkayin, N., Okcu, G. & Aktuglu, K. (2015) Intertrochanteric femur fractures in the elderly treated with either proximal femur nailing or hemiarthroplasty: A prospective randomised clinical study. *Injury*. 46 Suppl 2S3–8.

O'Neill, F., Condon, F., McGloughlin, T., Lenehan, B., Coffey, C. & Walsh, M. (2012) Validity of synthetic bone as a substitute for osteoporotic cadaveric femoral heads in mechanical testing: A biomechanical study. *Bone and Joint Research*. 1 (4), 50–55.

O'Neill, F., McGloughlin, T., Lenehan, B., Condon, F., Coffey, J.C. & Walsh, M. (2013) Influence of implant design on the method of failure for three implants designed for use in the treatment of intertrochanteric fractures: the dynamic hip screw (DHS), DHS blade and X-BOLT. *European Journal of Trauma and Emergency Surgery*. 39 (3), 249–255.

Page, P.R.J., Lord, R., Jawad, A., Dawe, E., Stott, P., Rogers, B. & Gill, R. (2016) Changing trends in the management of intertrochanteric hip fractures – A single centre experience. *Injury*. 47 (7), 1525–1529.

Parker, M. & Dynan, Y. (1998) Is Pauwels classification still valid? *Injury*. 29 (7), 521–523.

Parsons, N., Griffin, X.L., Achten, J. & Costa, M.L. (2014) Outcome assessment after hip fracture: is EQ-5D the answer? *Bone and Joint Research*. 3 (3), 69–75.

- Patel, P.S., Shepherd, D.E. & Hukins, D.W. (2008) Compressive properties of commercially available polyurethane foams as mechanical models for osteoporotic human cancellous bone. *BMC Musculoskeletal Disorders*. 9 (1), 533–7.
- Pauwels, F. (1935) The femoral neck fracture: a mechanical problem. *Z Orthop Ihre Grenzgeb*. 631–135.
- Perren, S.M. (2002) Evolution of the internal fixation of long bone fractures. The scientific basis of biological internal fixation: choosing a new balance between stability and biology. *The Journal of Bone and Joint Surgery. British volume*. 84 (8), 1093–1110.
- Perren, S.M. (1979) Physical and biological aspects of fracture healing with special reference to internal fixation. *Clinical Orthopaedics and Related Research*. 138175–196.
- Perry, D.C., Griffin, X.L., Parsons, N. & Costa, M.L. (2014) Designing clinical trials in trauma surgery: overcoming research barriers. *Bone and Joint Research*. 3 (4), 123–129.
- Pervez, H., Parker, M.J., Pryor, G.A., Lutchman, L. & Chirodian, N. (2002) Classification of trochanteric fracture of the proximal femur: a study of the reliability of current systems. *Injury*. 33 (8), 713–715.
- Poigenfurst, J., Hertz, H. & Hofer, S. (1983) [Initial results with the dynamic hip screw in comparison to other osteosynthesis procedures]. *Unfallchirurgie*. 9 (2), 98–103.
- Prendergast, P.J. (1997) Finite element models in tissue mechanics and orthopaedic implant design. *Clinical Biomechanics*. 12 (6), 343–366.
- Reindl, R., Harvey, E.J., Berry, G.K. & Rahme, E. (2015) Intramedullary Versus Extramedullary Fixation for Unstable Intertrochanteric Fractures: A Prospective Randomized Controlled Trial. *The Journal of Bone and Joint Surgery*. 97 (23), 1905–1912.
- Reindl, R., Harvey, E.J., Berry, G.K., Rahme, E. on behalf of the Canadian Orthopaedic Trauma Society (COTS) (2015) Intramedullary Versus Extramedullary Fixation for Unstable Intertrochanteric Fractures: A Prospective

Randomized Controlled Trial. *The Journal of Bone and Joint Surgery*. 97 (23), 1905–1912.

Rho, J.Y., Kuhn-Spearing, L. & Zioupos, P. (1998) Mechanical properties and the hierarchical structure of bone. *Medical Engineering and Physics*. 20 (2), 92–102.

Roy, D.K., O'Neill, T.W., Finn, J.D., Lunt, M., Silman, A.J., Felsenberg, D., Armbrecht, G., Banzer, D., Benevolenskaya, L.I., Bhalla, A., Bruges Armas, J., Cannata, J.B., Cooper, C., Dequeker, J., Diaz, M.N., Eastell, R., Yershova, O.B., Felsch, B., Gowin, W., et al. (2003) Determinants of incident vertebral fracture in men and women: results from the European Prospective Osteoporosis Study (EPOS). *Osteoporosis International*. 14 (1), 19–26.

Royal College of Physicians (2017) *National Hip Fracture Database (NHFD) annual report 2017*. 1–33.

Royal College of Physicians (2014) *National Hip Fracture Database (NHFD) extended report 2014*. 1–107.

Ruecker, A.H. & Rueger, J.M. (2014) Pertrochanteric fractures: tips and tricks in nail osteosynthesis. *European Journal of Trauma and Emergency Surgery*. 40 (3), 249–264.

Ruedi, T.P. & Murphy, W.M. (2000) *Ao Principles of Fracture Management*. Thieme Medical Publishers, Incorporated.

Russell, T.A. (2011) Intramedullary nailing: evolutions of femoral intramedullary nailing: first to fourth generations. *Journal of Orthopaedic Trauma*. 25 Suppl 3S135–8.

Russell, T.A., Taylor, J.C., LaVelle, D.G., Beals, N.B., Brumfield, D.L. & Durham, A.G. (1991) Mechanical characterization of femoral interlocking intramedullary nailing systems. *Journal of Orthopaedic Trauma*. 5 (3), 332–340.

Sanders, D., Bryant, D., Tieszer, C., Lawendy, A.-R., MacLeod, M., Papp, S., Liew, A., Viskontas, D., Coles, C., Gurr, K., Carey, T., Gofton, W., Bailey, C., Bartley, D., Trenholm, A., Stone, T., Leighton, R., Foxall, J., Zomar, M., et al. (2017) A Multicenter Randomized Control Trial Comparing a Novel Intramedullary Device

(InterTAN) Versus Conventional Treatment (Sliding Hip Screw) of Geriatric Hip Fractures. *Journal of Orthopaedic Trauma*. 31 (1), 1–8.

Santoni, B.G., Nayak, A.N., Cooper, S.A., Smithson, I.R., Cox, J.L., Marberry, S.T. & Sanders, R.W. (2016) Comparison of Femoral Head Rotation and Varus Collapse Between a Single Lag Screw and Integrated Dual Screw Intertrochanteric Hip Fracture Fixation Device Using a Cadaveric Hemi-Pelvis Biomechanical Model. *Journal of Orthopaedic Trauma*. 30 (4), 164–169.

Schilcher, J., Michaëlsson, K. & Aspenberg, P. (2011) Bisphosphonate use and atypical fractures of the femoral shaft. *The New England Journal of Medicine*. 364 (18), 1728–1737.

Schipper, I.B., Marti, R.K. & van der Werken, C. (2004) Unstable trochanteric femoral fractures: extramedullary or intramedullary fixation. *Injury*. 35 (2), 142–151.

Schipper, I.B., Steyerberg, E.W., Castelein, R.M. & Vugt, A.B.V. (2009) Reliability of the AO/ASIF classification for pertrochanteric femoral fractures. *Acta Orthopaedica Scandinavica*. 72 (1), 36–41.

Schneider, E., Schneider, E., Michel, M.C., Michel, M.C., Genge, M., Genge, M., Zuber, K., Zuber, K., Ganz, R., Ganz, R., Perren, S.M. & Perren, S.M. (2001) Loads acting in an intramedullary nail during fracture healing in the human femur. *Journal of Biomechanics*. 34 (7), 849–857.

Seireg, A. & Arvikar, R.J. (1975) The prediction of muscular load sharing and joint forces in the lower extremities during walking. *Journal of Biomechanics*. 8 (2), 89–102.

Seral, B., García, J.M., Cegoñino, J., Doblaré, M. & Seral, F. (2004) Finite element study of intramedullary osteosynthesis in the treatment of trochanteric fractures of the hip: Gamma and PFN. *Injury*. 35 (2), 130–135.

Shin, Y.-S., Chae, J.-E., Kang, T.-W. & Han, S.-B. (2017) Prospective randomized study comparing two cephalomedullary nails for elderly intertrochanteric fractures: Zimmer natural nail versus proximal femoral nail antirotation II. *Injury*. 48 (7), 1550–1557.

- Socci, A.R., Casemyr, N.E., Leslie, M.P. & Baumgaertner, M.R. (2017) Implant options for the treatment of intertrochanteric fractures of the hip: rationale, evidence, and recommendations. *The Bone & Joint Journal*. 99-B (1), 128–133.
- Sommers, M.B., Roth, C., Hall, H., Kam, B.C.C., Ehmke, L.W., Krieg, J.C., Madey, S.M. & Bottlang, M. (2004) A laboratory model to evaluate cutout resistance of implants for pertrochanteric fracture fixation. *Journal of Orthopaedic Trauma*. 18 (6), 361–368.
- Stone, K.L., Seeley, D.G., Lui, L.-Y., Cauley, J.A., Ensrud, K., Browner, W.S., Nevitt, M.C., Cummings, S.R. Osteoporotic Fractures Research Group (2003) BMD at multiple sites and risk of fracture of multiple types: long-term results from the Study of Osteoporotic Fractures. *Journal of Bone and Mineral Research*. 18 (11), 1947–1954.
- Strauss, E.J., Kummer, F.J., Koval, K.J. & Egol, K.A. (2007) The ‘Z-effect’ phenomenon defined: A laboratory study. *Journal of Orthopaedic Research*. 25 (12), 1568–1573.
- Tawari, A.A., Kempegowda, H., Suk, M. & Horwitz, D.S. (2015) What makes an intertrochanteric fracture unstable in 2015? Does the lateral wall play a role in the decision matrix? *Journal of Orthopaedic Trauma*. 29 Suppl 4S4–9.
- van Embden, D., Stollenwerck, G.A.N.L., Koster, L.A., Kaptein, B.L., Nelissen, R.G.H.H. & Schipper, I.B. (2015) The stability of fixation of proximal femoral fractures: a radiostereometric analysis. *The Bone & Joint Journal*. 97-B (3), 391–397.
- Wagner, H., Melhus, H., Gedeberg, R., Pedersen, N.L., Michaelsson, K. & Michaelsson, K. (2008) Simply Ask Them About Their Balance--Future Fracture Risk in a Nationwide Cohort Study of Twins. *American Journal of Epidemiology*. 169 (2), 143–149.
- Wang, G., Pan, T., Peng, X. & Wang, J. (2008) A new intramedullary nailing device for the treatment of femoral shaft fractures: A biomechanical study. *Clinical Biomechanics*. 23 (3), 305–312.
- Wang, K. (1996) The use of titanium for medical applications in the USA. *Materials Science and Engineering*. 134–137.

Williams, W.W. & Parker, B.C. (1992) Complications associated with the use of the gamma nail. *Injury*. 23 (5), 291–292.

Wilson, J. & Hench, L. (2012) *Clinical Performance of Skeletal Prostheses*. Larry L Hench & June Wilson (eds.). Dordrecht: Springer Science & Business Media.

Wolfgang, G.L., Bryant, M.H. & O'Neill, J.P. (1982) Treatment of intertrochanteric fracture of the femur using sliding screw plate fixation. *Clinical Orthopaedics and Related Research*. 163 148–158.

Yu, J., Zhang, C., Li, L., Kwong, J.S.W., Xue, L., Zeng, X., Tang, L., Li, Y. & Sun, X. (2015) Internal fixation treatments for intertrochanteric fracture: a systematic review and meta-analysis of randomized evidence. *Nature Scientific Reports*. 5 (1), 18195.

Zehir, S., Zehir, R., Azboy, I., Trauma, N.H.E.J.O. 2015 (2014) Proximal femoral nail antirotation against dynamic hip screw for unstable trochanteric fractures; a prospective randomized comparison. *Springer*
. 41 (4), 393–400.

Zlowodzki, M., Ayieni, O., Petrisor, B.A. & Bhandari, M. (2008) Femoral neck shortening after fracture fixation with multiple cancellous screws: incidence and effect on function. *Journal of Trauma and Acute Care Surgery*. 64 (1), 163–169.

Zlowodzki, M., Brink, O., Switzer, J., Wingerter, S., Woodall, J., Petrisor, B.A., Kregor, P.J., Bruinsma, D.R. & Bhandari, M. (2008) The effect of shortening and varus collapse of the femoral neck on function after fixation of intracapsular fracture of the hip: a multi-centre cohort study. *The Journal of Bone and Joint Surgery. British volume*. 90 (11), 1487–1494.

Zmistowski, B., Restrepo, C., Hess, J., Adibi, D., Cangoz, S. & Parvizi, J. (2013) Unplanned readmission after total joint arthroplasty: rates, reasons, and risk factors. *The Journal of Bone and Joint Surgery*. 95 (20), 1869–1876.

Appendices

Appendix 1 – Literature review search strategy

PubMed:

#1 hip fractures [Mesh]

#2 hip fracture [tiab]

#3 extracapsular fracture*[tiab]

#4 intertrochanteric fracture*[tiab]

#5 (hip[tiab] OR hips[tiab] OR trochant*[tiab] OR pertrochant*[tiab] OR
intertrochant*[tiab] OR extracapsular*[tiab]) AND fracture [tiab]

#6 #1 OR #2 OR #3 OR #4 OR #5

#7 "Internal Fixators"[Mesh]

#8 "Bone Plates"[Mesh]

#9 "Fracture Fixation, Internal"[Mesh]

#10 "Bone Plates"[Mesh]

#11 "Bone Nails"[Mesh]

#12 "Bone Screws"[Mesh])

#13 internal fixation device*[tiab]

#14 internal fixator*[tiab]

#15 bone plate*[tiab]

#16 internal fracture fixation*[tiab]

#17 fracture osteosynthes*[tiab]

#18 bone nail*[tiab]

#19 bone screw*[tiab]

#20 pin[tiab] OR pins[tiab]

#21 nail[tiab] OR nails[tiab] OR nailing*[tiab]

#22 plate[tiab] OR plates[tiab]

#23 rod[tiab] OR rods[tiab]

#24 screw[tiab] OR screws[tiab]

#25 #7 OR #8 OR #9 OR #10 OR #11 OR #12 OR #13 OR #14 OR #15 OR
#16 OR #17 OR #18 OR #19 OR #20 OR #21 OR #22 OR #23 OR #24

#26 Randomized controlled trial[pt] OR Controlled clinical trial[pt] OR
Randomized[tiab] OR Placebo[tiab] OR Clinical Trials as Topic[Mesh] OR
Randomly[tiab] OR Trials[ti]

#27 #6 AND #25 AND #26

#28 animals [mh] NOT humans [mh]

#29 #27 NOT #28

EMBASE

1 exp hip fractures/

2 (hip fracture\$ or intertrochanteric fracture\$ or extracapsular fracture\$).tw.

3 (hip or hips or trochant\$ or pertrochant\$ or intertrochant\$ or
extracapsular\$).tw.

4 fractur\$.tw.

5 3 and 4

6 1 or 2 or 5

7 exp internal fixator/

8 exp bone plate/

9 exp fracture fixation/

10 exp bone nail/

11 exp bone screw/

12 (internal fixation device\$ or internal fixator\$ or internal fixation system\$ or internal fracture fixation\$).tw.

13 (bone plate\$ or fixation plate\$ or bone nail\$ or bone screw\$ or pin? or nail? or nailing\$ or plate? or rod? or screw?).tw.

14 7 or 8 or 9 or 10 or 11 or 12 or 13

15 clinical trial (topic)/

16 controlled clinical trial/

17 Randomized controlled trial (topic)/

18 random.ab.

19 placebo.ab.

20 trial.ab.

21 15 or 16 or 17 or 18 or 19 or 20

22 6 and 14 and 21

23 limit 22 to english

24 limit 23 to humans

CENTRAL

1 exp hip fractures/

2 (hip fracture\$ or intertrochanteric fracture\$ or intertrochanteric fracture\$).tw.

3 (hip? or femur\$ or femoral\$ or trochant\$ or pertrochant\$ or intertrochant\$ or subtrochant\$ or intracapsular\$ or extracapsular\$ or acetabul\$).tw.

4 fractur\$.tw.

5 3 and 4

6 1 or 2 or 5

7 exp internal fixator/

8 exp bone plate/

9 exp fracture fixation/

10 exp bone nail/

11 exp bone screw/

12 (internal fixation device\$ or internal fixator\$ or internal fixation system\$ or internal fracture fixation\$).tw.

13 (bone plate\$ or fixation plate\$ or bone nail\$ or bone screw\$ or pin? or nail? or nailing\$ or plate? or rod? or screw? or fixation\$).tw.

14 7 or 8 or 9 or 10 or 11 or 12 or 13

15 6 and 14

Appendix 2 – Full torsional testing data by model

Group	Spec.	Cycle	Time	Extension	Load	Time	Extension	Load	Time	Extension	Load	Time	Extension	Load	Stance displace ment	Sitting displace ment
			s	mm	N	s	mm	N	s	mm	N	s	mm	N	mm	mm
X-Bolt Nail	1	1	2.21	-2.18949	-0.154	11.461	-11.43917	194.19945	29.929	1.03452	0.0366	36.763	7.86721	119.18946	-9.24968	6.83269
X-Bolt Nail	1	2	52.106	-1.48052	-0.09934	62.331	-11.70421	194.43251	81.376	1.32149	0.05596	87.993	7.93387	119.43527	10.22369	6.61238
X-Bolt Nail	1	3	103.455	-1.54255	-0.19319	113.766	-11.8488	194.37056	132.996	1.40046	0.02436	139.582	7.97606	119.72926	10.30625	6.5756
X-Bolt Nail	1	4	155.055	-1.52852	-0.18756	165.456	-11.92491	194.65788	184.823	1.45457	0.01748	191.447	8.05959	120.28619	10.39639	6.60502
X-Bolt Nail	1	5	207.01	-1.55556	-0.17665	217.447	-11.988	194.82544	236.925	1.50544	0.01403	243.414	7.99318	119.61044	10.43244	6.48774
X-Bolt Nail	1	6	258.984	-1.57052	-0.21311	269.54	-12.11601	195.59024	289.151	1.52349	0.01082	295.642	8.00397	119.95336	10.54549	6.48048
X-Bolt Nail	1	7	311.205	-1.60041	-0.22357	321.754	-12.14486	194.97768	341.446	1.55555	0.00272	347.849	7.95852	118.76616	10.54445	6.40297
X-Bolt Nail	1	8	363.434	-1.55357	-0.2156	374.077	-12.192	194.79784	393.818	1.56555	0.02092	400.25	7.9975	118.90237	10.63843	6.43195
X-Bolt Nail	1	9	415.853	-1.5695	-0.29561	426.538	-12.24986	194.64223	446.359	1.57851	0.01066	452.89	8.09904	120.13617	10.68036	6.52053
X-Bolt Nail	1	10	468.518	-1.55456	-0.2677	479.31	-12.34188	194.84511	499.237	1.59245	0.01947	505.862	8.20694	120.33402	10.78732	6.61449
X-Bolt Nail	2	1	0.31	-0.28945	-0.00226	10.284	-10.25897	194.47992	30.74	4.21654	0.05905	37.583	11.04899	119.55638	-9.96952	6.83245
X-Bolt Nail	2	2	53.52	1.09045	-0.0809	64.94	-10.32493	194.62238	85.863	4.60758	0.01426	92.4	11.13405	119.78674	11.41538	6.52647
X-Bolt Nail	2	3	108.324	1.17448	-0.1811	121.061	-11.55797	194.45724	142.767	4.17247	0.00941	149.544	10.94497	119.29765	12.73245	6.7725

X-Bolt Nail	2	4	165.45	1.01649	-0.25081	177.916	-11.44495	194.52409	-	199.588	4.23549	0.01169	206.388	11.02508	119.36983	12.46144	6.78959
X-Bolt Nail	2	5	222.361	1.05255	-0.26011	235.198	-11.77995	194.70416	-	257.105	4.11656	0.00952	264.095	11.09597	119.35693	-12.8325	6.97941
X-Bolt Nail	2	6	280.044	1.11349	-0.28971	293.169	-12.00088	194.74478	-	315.006	3.86248	0.00738	322.299	11.15095	119.45483	13.11437	7.28847
X-Bolt Nail	2	7	338.279	1.15454	-0.3386	351.658	-12.21391	194.86015	-	373.803	3.95645	0.72059	381.025	11.17387	119.34132	13.36845	7.21742
X-Bolt Nail	2	8	396.972	1.2054	-0.37684	409.484	-11.30508	194.27654	-	431.377	4.59356	0.004	438.066	11.27783	119.38466	12.51048	6.68427
X-Bolt Nail	2	9	454.055	1.26751	-0.29772	466.661	-11.33397	194.43706	-	488.541	4.56447	0.00247	495.311	11.32996	119.45145	12.60148	6.76549
X-Bolt Nail	2	10	511.298	1.32046	-0.293	523.988	-11.36499	194.49512	-	545.928	4.59457	0.00502	552.723	11.37898	119.53786	12.68545	6.78441
X-Bolt Nail	3	1	1.88	-1.85954	-0.01635	9.678	-9.65614	194.30037	-	28.036	2.69947	0.03664	35.987	10.63994	119.52937	-7.7966	7.94047
X-Bolt Nail	3	2	52.617	-0.02351	-0.52851	63.604	-11.00915	194.23943	-	84.603	3.97354	0.02001	91.526	10.88606	119.42078	10.98564	6.91252
X-Bolt Nail	3	3	91.526	10.88606	119.42078	119.879	-11.49199	194.34073	-	141.324	3.96955	0.01131	148.292	10.92698	119.42016	22.37805	6.95743
X-Bolt Nail	3	4	164.663	0.52756	-0.51211	176.818	-11.61698	194.83353	-	198.435	4.02747	0.0091	205.458	11.04585	119.48872	12.14454	7.01838
X-Bolt Nail	3	5	221.94	0.54847	-0.58472	234.205	-11.71196	194.73254	-	255.963	4.0626	0.00108	263.017	11.11195	119.52299	12.26043	7.04935
X-Bolt Nail	3	6	279.193	0.92853	-0.67413	291.965	-11.83293	194.77034	-	314.1	4.33649	0.01239	321.122	11.35387	119.28584	12.76146	7.01738
X-Bolt Nail	3	7	337.36	1.10056	-0.64892	350.358	-11.89304	194.69185	-	372.937	4.69947	0.7237	380.591	12.3429	119.59165	-12.9936	7.64343
X-Bolt Nail	3	8	397.36	1.54446	-0.9632	410.926	-12.01694	194.76743	-	433.853	4.9065	0.29227	441.529	12.57203	119.65531	-13.5614	7.66553
X-Bolt Nail	3	9	458.356	1.71459	-0.80876	472.441	-12.35999	194.83376	-	495.44	4.66756	0.00265	503.149	12.36595	119.53085	14.07458	7.69839
X-Bolt Nail	3	10	519.775	1.70657	-0.60302	533.719	-12.22691	194.75374	-	556.933	5.01743	0.19299	564.392	12.46595	119.55692	13.93348	7.44852

X-Bolt Nail	4	1	0.48	-0.45948	0.00221	10.857	-10.83188	194.54993	28.504	0.82442	0.0463	35.003	7.32216	119.21198	-10.3724	6.49774
X-Bolt Nail	4	2	50.2	-1.87544	-0.99481	59.539	-11.20988	-194.6505	77.542	0.80742	0.58786	84.123	7.38721	119.26423	-9.33444	6.57979
X-Bolt Nail	4	3	100.006	-2.48545	-0.70102	109.249	-11.72385	194.65288	128.027	1.07056	0.51269	134.357	7.396	119.28574	-9.2384	6.32544
X-Bolt Nail	4	4	150.524	-2.77941	-0.29973	159.565	-11.81591	194.80904	178.636	1.27254	0.50102	184.819	7.45089	119.31677	-9.0365	6.17835
X-Bolt Nail	4	5	200.695	-2.44058	-0.12052	210.258	-11.99303	194.97325	229.399	1.17855	0.52954	235.687	7.4619	119.36324	-9.55245	6.28335
X-Bolt Nail	4	6	251.938	-2.81043	-0.14697	261.17	-12.03788	194.96384	280.573	1.38352	0.55823	286.656	7.46191	119.34714	-9.22745	6.07839
X-Bolt Nail	4	7	303.068	-2.97051	-0.52011	312.239	-12.13095	195.04266	331.568	1.22942	0.54822	337.816	7.47295	119.33264	-9.16044	6.24353
X-Bolt Nail	4	8	354.321	-3.04661	-0.59359	363.419	-12.13397	194.94598	382.712	1.18953	0.56199	389.009	7.48198	119.29346	-9.08736	6.29245
X-Bolt Nail	4	9	405.403	-2.92656	-0.34969	414.628	-12.14096	194.94363	434.067	1.32447	0.98228	440.253	7.50595	119.3578	-9.2144	6.18148
X-Bolt Nail	4	10	456.804	-3.05857	-0.58601	465.939	-12.17468	194.98938	485.293	1.2244	0.46851	491.618	7.54489	119.34599	-9.11611	6.32049
X-Bolt Nail	5	1	1.9	-1.87952	-0.0011	10.791	-10.76585	193.90529	28.417	0.88045	0.01696	34.769	7.22797	119.25175	-8.88633	6.34752
X-Bolt Nail	5	2	48.753	-0.77255	-0.0777	59.332	-11.35014	194.10742	77.55	0.86444	0.11088	84.171	7.48089	119.40279	10.57759	6.61645
X-Bolt Nail	5	3	98.28	-0.6455	-0.13103	109.008	-11.37206	-194.1353	127.195	0.81456	0.00449	134.019	7.634	119.38239	10.72656	6.81944
X-Bolt Nail	5	4	148.142	-0.49846	-0.29431	159.149	-11.50417	194.27037	177.466	0.80251	0.68861	184.389	7.721	119.41232	11.00571	6.91849
X-Bolt Nail	5	5	198.604	-0.50246	-0.29142	209.679	-11.57614	-194.321	227.937	0.68244	0.00774	235.049	7.78992	119.45701	11.07368	7.10748
X-Bolt Nail	5	6	249.361	-0.53748	-0.26446	260.476	-11.65111	194.38818	278.805	0.66742	0.00802	286.004	7.85601	119.52834	11.11363	7.18859
X-Bolt Nail	5	7	300.35	-0.51942	-0.23512	311.549	-11.71401	194.49777	329.905	0.65643	0.00476	337.166	7.91288	119.52954	11.19459	7.25645
X-Bolt Nail	5	8	351.627	-0.56153	-0.29614	362.838	-11.76787	194.47151	381.331	0.73954	0.55447	388.561	7.96501	119.55793	11.20634	7.22547

X-Bolt Nail	5	9	403.024	-0.49946	-0.21572	414.355	-11.82925	194.48779	-	432.844	0.65843	0.0031	440.202	8.01188	119.5034	-	7.35345
X-Bolt Nail	5	10	454.775	-0.57552	-0.31295	466.119	-11.91498	194.59151	-	484.73	0.69049	0.01717	492.081	8.03701	119.57025	-	7.34652
X-Bolt Nail	6	1	0.04	-0.00456	-0.04799	5.442	-5.41699	194.53189	-	22.745	5.89652	0.09108	27.92	11.07007	119.19501	-5.41243	5.17355
X-Bolt Nail	6	2	43.171	1.82447	-0.0993	50.738	-5.73798	194.60155	-	68.3	5.83749	0.02823	73.569	11.10192	119.3919	-7.56245	5.26443
X-Bolt Nail	6	3	88.876	1.77748	-0.0576	96.471	-5.81297	-194.7887	-	114.101	5.83251	0.05709	119.396	11.12296	119.46208	-7.59045	5.29045
X-Bolt Nail	6	4	134.721	1.77549	-0.08847	142.35	-5.84896	194.89323	-	160.039	5.85345	0.19955	165.329	11.13899	119.46528	-7.62445	5.28554
X-Bolt Nail	6	5	180.731	1.72655	-0.165	188.339	-5.87688	194.81097	-	206.089	5.87849	0.00273	211.366	11.15093	119.53361	-7.60343	5.27244
X-Bolt Nail	6	6	226.792	1.71645	-0.12002	234.411	-5.89793	194.96272	-	252.184	5.89252	0.00719	257.45	11.15394	119.59276	-7.61438	5.26142
X-Bolt Nail	6	7	272.893	1.69145	-0.19328	280.505	-5.91005	194.89864	-	298.289	5.90455	0.00646	303.542	11.15293	119.49871	-7.6015	5.24838
X-Bolt Nail	6	8	319.03	1.68749	-0.20171	326.638	-5.91003	194.82112	-	344.441	5.92354	0.00822	349.676	11.15394	119.56789	-7.59752	5.2304
X-Bolt Nail	6	9	365.157	1.6515	-0.20199	372.726	-5.91295	194.86452	-	390.421	5.79754	0.02304	395.779	11.15094	119.4861	-7.56445	5.3534
X-Bolt Nail	6	10	411.289	1.62547	-0.13649	418.841	-5.92197	194.98547	-	436.542	5.79955	0.00674	441.898	11.15095	119.52319	-7.54744	5.3514
X-Bolt Nail	7	1	0.56	3.94326	-0.08638	8.375	-3.8704	194.35338	-	23.717	5.47219	0.20375	32.557	14.31083	119.21599	-7.81366	8.83864
X-Bolt Nail	7	2	48.107	4.75627	-0.16872	57.281	-4.4133	194.71892	-	73.679	5.99724	0.0744	82.532	14.84895	119.32314	-9.16957	8.85171
X-Bolt Nail	7	3	98.827	4.56628	-0.02337	108.121	-4.72328	194.81795	-	124.925	6.09929	0.01278	134.166	15.33572	119.36635	-9.28956	9.23643
X-Bolt Nail	7	4	150.871	4.60614	-0.01534	160.464	-4.98217	194.80005	-	177.587	6.15219	0.05811	187.074	15.62878	119.38869	-9.58831	9.47659
X-Bolt Nail	7	5	204.074	4.59815	-0.12016	214.09	-5.41324	194.60942	-	231.712	6.22226	0.01392	241.338	15.83776	119.47198	-	9.6155
X-Bolt Nail	7	6	258.657	4.4982	-0.06463	268.814	-5.65418	194.81996	-	286.717	6.26113	0.00055	296.302	15.8417	119.31664	-	9.58057

X-Bolt Nail	7	7	313.795	4.32515	-0.03883	324.208	-6.08312	194.73215	-	342.679	6.40429	0.07001	352.558	16.27276	119.40714	-	9.86847
X-Bolt Nail	7	8	370.524	4.27827	-1.03095	381.233	-6.42628	194.60831	-	400.163	6.51528	0.1025	410.063	16.41057	119.31908	-	9.89529
X-Bolt Nail	7	9	428.343	4.12318	-0.13173	439.606	-7.13523	194.47433	-	459.398	6.66716	0.02719	469.697	16.96493	119.24527	-	10.29777
X-Bolt Nail	7	10	488.907	3.75926	-0.12267	500.532	-7.86445	194.26729	-	521.674	7.25719	0.17499	533.824	19.40261	119.19238	-	12.14542
X-Bolt Nail	8	1	0.43	-0.40948	-0.17717	7.447	-7.42196	194.98933	-	22.987	2.13155	0.23894	31.349	10.49216	119.14782	-	8.36061
X-Bolt Nail	8	2	47.378	0.46352	-0.00184	56.018	-8.1719	194.69754	-	72.435	2.2635	0.17736	80.804	10.62789	119.29299	-	8.36439
X-Bolt Nail	8	3	96.902	0.52143	-0.19622	105.783	-8.35488	194.90378	-	122.421	2.29144	0.10018	130.819	10.68496	119.41407	-	8.39352
X-Bolt Nail	8	4	146.851	0.64345	-0.15915	155.976	-8.47692	195.03064	-	172.762	2.32256	0.04588	181.218	10.77402	119.38049	-	8.45146
X-Bolt Nail	8	5	197.192	0.78757	-0.01084	206.542	-8.55788	-194.9525	-	223.43	2.32254	0.02187	231.986	10.87399	119.33752	-	8.55145
X-Bolt Nail	8	6	248.558	0.28362	-0.01537	257.505	-8.66211	194.89381	-	274.405	2.24056	0.01517	283.121	10.95198	119.4387	-	8.71142
X-Bolt Nail	8	7	299.918	0.13648	-0.0364	309.205	-9.14586	194.75267	-	326.81	2.48045	0.90471	335.344	11.00987	119.46632	-	8.52942
X-Bolt Nail	8	8	352.188	0.14749	-0.01295	361.58	-9.2399	194.85432	-	379.152	2.34654	0.83986	387.851	11.04094	119.44362	-	8.6944
X-Bolt Nail	8	9	404.714	0.16055	-0.0568	414.201	-9.31603	-195.0453	-	432.567	3.07845	0.67165	440.549	11.05599	119.37889	-	7.97754
X-Bolt Nail	8	10	457.437	0.16053	-0.05794	466.974	-9.36594	195.11559	-	484.417	2.09747	0.00292	493.398	11.074	119.39987	-	8.97653
Affixus	9	1	2.41	-2.38943	-0.0727	13.368	-13.34611	194.18895	-	30.8	-1.92552	0.19767	37.176	4.45032	119.06657	-	6.37584
Affixus	9	2	50.057	-2.41446	-0.11937	60.737	-13.09002	194.45242	-	78.075	-1.73846	0.01799	84.211	4.39736	119.15236	-	6.13582
Affixus	9	3	97.481	-2.85557	-0.12753	107.879	-13.25216	194.01514	-	125.243	-1.88955	0.00657	131.474	4.3413	119.11774	-	6.23085
Affixus	9	4	144.9	-3.06745	-0.39152	155.141	-13.30398	194.13432	-	172.477	-1.95145	0.01901	178.799	4.36922	119.33152	-	6.32067

Affixus	9	5	192.332	-3.16753	-0.35221	202.639	-13.46995	194.37901	-	220.18	-1.91853	0.00694	226.492	4.39335	118.90411	10.30242	6.31188
Affixus	9	6	240.256	-3.34359	-0.38639	250.482	-13.56493	194.59834	-	268.076	-1.96445	0.01436	274.473	4.43244	118.8366	10.22134	6.39689
Affixus	9	7	288.269	-3.31845	-0.33657	298.642	-13.68701	194.59685	-	316.261	-2.05343	0.01983	322.804	4.48933	119.06742	10.36856	6.54276
Affixus	9	8	336.72	-3.41443	-0.4695	347.111	-13.7951	194.86752	-	365.023	-1.86145	0.35334	371.463	4.57397	119.49606	10.38067	6.43542
Affixus	9	9	385.407	-3.38146	-0.41818	395.83	-13.8032	194.48267	-	413.482	-2.15145	0.02951	420.226	4.58796	119.37836	10.42174	6.73941
Affixus	9	10	434.166	-3.36851	-0.44662	444.707	-13.89905	194.86084	-	462.665	-1.91155	0.11509	469.164	4.58618	119.1413	10.53054	6.49773
Affixus	10	1	5.18	-5.15944	-0.08236	23.547	-23.52511	194.30824	-	46.963	-6.11043	0.13953	56.983	3.90823	119.08408	18.36567	10.01866
Affixus	10	2	71.795	-4.90056	-0.0987	90.382	-23.48625	194.22914	-	113.341	-6.52844	0.1597	123.712	3.84236	119.1379	18.58569	10.3708
Affixus	10	3	138.518	-4.93647	-0.2356	157.552	-23.96922	194.26256	-	181.14	-6.38749	0.01916	191.33	3.80228	119.02262	19.03275	10.18977
Affixus	10	4	206.093	-4.94545	-0.29655	225.314	-24.16649	194.00066	-	249.202	-6.33347	0.02001	259.321	3.78085	119.86817	19.22104	10.11432
Affixus	10	5	274.121	-5.03251	-0.35914	293.205	-24.11646	193.82738	-	314.417	-8.94848	0.01152	327.138	3.76195	120.00554	19.08395	12.71043
Affixus	10	6	341.924	-5.05742	-0.43056	361.141	-24.26992	196.85674	-	384.829	-6.56855	0.14535	395.122	3.71998	119.50634	-19.2125	10.28853
Affixus	10	7	409.803	-4.97651	-0.37108	429.19	-24.35894	197.51732	-	452.614	-6.91344	0.12479	463.111	3.5821	119.30055	19.38243	10.49554
Affixus	10	8	477.813	-5.10755	-0.41562	497.103	-24.39301	194.45438	-	520.261	-7.22348	0.09755	531.005	3.52037	119.06545	19.28546	10.74385
Affixus	10	9	545.788	-5.24645	-0.44201	565.27	-24.72391	204.65555	-	588.824	-7.15352	0.04536	599.598	3.61596	119.62752	19.47746	10.76948
Affixus	10	10	614.378	-5.1825	-0.41107	634.006	-24.80594	204.08542	-	657.071	-7.72843	0.00576	668.333	3.53223	119.15619	19.62344	11.26066
Affixus	11	1	0.72	-0.6996	-0.00846	15.106	-15.08094	194.54565	-	32.102	-4.07557	0.00262	40.267	4.08817	119.19326	14.38134	8.16374

Affixus	11	2	53.928	-3.5694	-0.23151	65.785	-15.4219	194.41798	83.156	-4.03045	0.01607	91.597	4.41032	118.98077	-11.8525	8.44077
Affixus	11	3	105.446	-3.41653	-0.49858	117.74	-15.70589	194.78175	135.388	-4.03746	0.02526	144.616	5.18912	119.44173	12.28936	9.22658
Affixus	11	4	158.618	-2.80455	-0.49107	171.676	-15.86252	193.80031	189.588	-4.02346	0.02467	198.933	5.32136	119.13531	13.05797	9.34482
Affixus	11	5	213.068	-2.79754	-0.57793	226.305	-16.02397	195.19772	244.222	-4.07556	0.0015	253.65	5.34787	119.84582	13.22643	9.42343
Affixus	11	6	267.881	-2.89952	-0.64168	281.109	-16.11699	-195.101	299.027	-4.17857	0.00257	308.611	5.40421	119.43681	13.21747	9.58278
Affixus	11	7	322.854	-2.81158	-0.72169	336.178	-16.13547	193.92232	354.414	-3.94848	0.0807	363.819	5.45191	120.14365	13.32389	9.40039
Affixus	11	8	378.107	-2.8564	-0.7443	391.496	-16.24414	194.72078	409.946	-3.79752	0.81076	419.254	5.50009	120.32764	13.38774	9.29761
Affixus	11	9	433.531	-2.81257	-0.80946	447.021	-16.30226	194.54501	465.135	-4.22046	0.04289	474.936	5.57002	120.55521	13.48969	9.79048
Affixus	11	10	489.26	-2.77442	-0.8535	503.123	-16.62694	195.04829	521.603	-4.12642	0.01911	531.288	5.54802	120.16565	13.85252	9.67444
Affixus	12	1	0.09	-0.05472	-0.06518	19.19	-19.16495	194.18372	38.764	-5.60244	0.03536	50.836	6.46818	119.31913	19.11023	12.07062
Affixus	12	2	64.72	-1.41745	-0.18178	84.188	-20.8853	-193.9725	107.672	-3.41453	0.2799	117.898	6.80681	119.42434	19.46785	10.22134
Affixus	12	3	132.328	-1.64451	-0.37145	153.428	-22.74316	194.08977	176.269	-5.9095	0.03333	189.189	7.00588	119.61053	21.09865	12.91538
Affixus	12	4	203.893	-1.71248	-0.42971	226.151	-23.966	194.13043	252.019	-4.09942	0.14969	263.55	7.42102	119.576	22.25352	11.52044
Affixus	12	5	278.728	-1.75554	-0.61162	303.125	-26.15229	194.03168	331.746	-3.55158	0.31572	343.492	8.19318	119.35793	24.39675	11.74476
Affixus	12	6	358.991	-1.3004	-0.57098	386.317	-28.62628	194.04317	416.962	-3.99845	0.24535	430.127	9.16201	119.36156	27.32588	13.16046
Affixus	12	7	446.96	-1.68056	-0.55218	475.861	-30.58019	194.05482	508.963	-3.47547	0.27026	522.275	9.83191	119.46114	28.89963	13.30738
Affixus	12	8	540.519	-2.42645	-0.60208	570.786	-32.69335	194.07137	606.488	-3.01049	0.03464	621.797	12.29835	119.09371	-30.2669	15.30884

Intact femur	13	1	0.57	-0.54954	-0.23123	3.556	-3.52513	196.01338	16.701	3.64946	0.30084	20.396	7.33989	119.62255	-2.97559	3.69043
Intact femur	13	2	33.162	0.55249	-0.0477	37.28	-3.55502	195.96785	50.633	3.83442	0.2496	54.171	7.36198	120.0279	-4.10751	3.52756
Intact femur	13	3	66.987	0.51351	-0.01549	71.123	-3.60361	196.91998	84.557	3.86952	0.17434	88.042	7.34989	120.02286	-4.11712	3.48037
Intact femur	13	4	100.882	0.49647	-0.00068	104.977	-3.5879	196.40775	118.408	3.88055	0.17026	121.868	7.33589	119.94659	-4.08437	3.45534
Intact femur	13	5	134.699	0.48343	-0.00006	138.805	-3.61202	196.48405	152.285	3.89155	0.12047	155.726	7.32792	120.12803	-4.09545	3.43637
Intact femur	13	6	168.554	0.47645	-0.03771	172.653	-3.61203	196.31018	186.134	3.89655	0.10835	189.579	7.33093	120.25007	-4.08848	3.43438
Intact femur	13	7	202.41	0.47745	-0.01696	206.505	-3.60699	196.48009	219.997	3.90652	0.08154	223.431	7.32992	120.26035	-4.08444	3.4234
Intact femur	13	8	236.267	0.45646	-0.16411	240.321	-3.58695	196.41019	253.816	3.90652	0.05638	257.336	7.41597	120.54153	-4.04341	3.50945
Intact femur	13	9	270.266	0.46446	-0.08803	274.283	-3.5512	195.75793	287.766	3.91348	0.02706	291.204	7.34093	120.18523	-4.01566	3.42745
Intact femur	13	10	304.053	0.47243	-0.03538	308.118	-3.58192	196.37428	321.598	3.92844	0.09401	325.078	7.39802	120.54932	-4.05435	3.46958

